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WALKING RESPONSES UNDER LUNAR AND LOW GRAVITY CONDITIONS

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BEHAVIORAL SCIENCES LABORATORY
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AEROSPACE MEDICAL DIVISION
AIR FORCE SYSTEMS COMMAND
WRIGHT-PATTERSON AIR FORCE BASE, OHIO

Project No. 7184, Task No. 718405

NOTICES

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". . They danced on the ceilings,

They danced on the walls,

At the Gandy Dancers' Ball."

From "The Gandy Dancers' Ball"
American Folk Ballad

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ABSTRACT

Previous walking studies are reviewed to determine methods of investigation. A description of normal walking is included for definition and as a basis for comparison of the low gravity gait. Walking behavior can be analyzed by motion methods or by force methods. Motion analysis is sufficient to quantitatively describe the low gravity gait, but a force analysis is needed to establish the reasons for the degradation of the walk. The construction of a force-measuring walkway is proposed, but was not completed. Alternatively, a motion picture (time-displacement) walking experiment was conducted at various artificial gravity levels in an aircraft flying parabolic trajectories. Two subjects walked fore and aft on a floor-marked distance scale. Successive lowering of the gravity level from 1.0 g to 0.1 g produced the following effects. The subjects maintained within 10 percent of their normal 1-g velocity to gravity levels of 0.25 g or below. The most consistent effect of the reduction of gravity was the increased swinging time of the leg, which varied inversely as the sixth root of the gravity level. The increasing swing-to-support ratio indicates the decreasing control of progression, and leads one to describe the low-gravity gait as "a fast walk in slow motion." The discontinuity in the performance curves in the region of 0.2 g tends to substantiate previous estimates of the lower gravity limit for walking.

PUBLICATION REVIEW

This technical documentary report is approved.

Walter F. Grether

WALTER F. GRETHER
Technical Director
Behavioral Sciences Laboratory

FOREWORD

This locomotion study was suggested by Captain John C. Simons, Crew Stations Branch, Human Engineering Division, Behavioral Sciences Laboratory, 6570th Aerospace Medical Research Laboratories, and fulfills a requirement under Project No. 7184, "Human Performance in Advanced Systems," Task No. 718405, "Design Criteria for Crew Stations in Advanced Systems."

The report was prepared by Captain Roberts in partial fulfillment of the requirements for the Master of Science Degree in Engineering at USAF Institute of Technology in 1962.

Special acknowledgment is made to Captain Simons and to Lieutenant Bucciarelli, thesis adviser, for their sincere interest, suggestions, and guidance throughout this study. Also, thanks are extended to Mr. R. E. Conover, Mr. V. F. Willis, Mr. E. C. Titterington, and Mr. F. C. Shera of the Data Systems Division, Directorate of Test Data, Deputy for Flight Test, for their assistance in the design of the strain gage instrumentation; and to Sergeants William Sears and Harold Espensen, Crew Stations Branch, for their assistance in all phases of the experimental tests; and to Mr. Charles E. Clauser, Anthropology Branch, Human Engineering Division, for his many suggestions and literature guidance.

I. Introduction

Subject and Purpose

The subject of this study is the walking performance of a man under the conditions of reduced gravity. The purpose is to objectively determine his capabilities and to quantitatively describe his performance.

Subject Background

Encumbered by a space suit and other life-support equipment, it is unlikely that the astronauts, like the railroad men of old, will be "dancing on the walls." That is, at least not voluntarily. But the questions, "Will he be able to walk?" and if so, "How well?" remain unanswered. In the near future the Apollo mission will require that the astronaut emerge from his landing vehicle and walk about on the surface of the moon. In fact, until lunar roving vehicles are available, the primary means of locomotion on the moon will be normal walking.

Lunar walking will be only the first of the requirements for man to walk in space. Duties aboard manned orbital space stations and exploration of the near planets will undoubtedly require man to walk in a variety of gravity fields. These requirements, coupled with the lack of existing subject knowledge, establish the need for this investigation. The basic assumption that it is desirable to learn as much as possible about man's capabilities before he is placed in an adverse environment provides the motivation for its continuance.

Scope

This study is concerned with the effect of the reduction of gravity on man's walking performance. The initial objectives of the study are as follows:

1. To describe in quantitative terms the walking performance that is possible at reduced gravity levels,
2. To determine the lower limit at which man can walk unaided, and
3. To determine the reasons for man's relative failure or success in walking.

This study is limited to man's capabilities in a "shirtsleeve" environment under the influence of gravity levels between 0.1 g and 1.0 g. No attempt is made to evaluate walking performance in the absence of gravity, that is, under purely "weightless" conditions. Emphasis is placed on describing the dynamics of the gait rather than on the biological phenomena within the body that produce this motion.

The approach is primarily experimental. Because of the complexity of walking, mathematical analysis is used to explain only certain facets of the gait rather than to derive the whole walking performance.

This investigation is of a preliminary nature and serves as a guideline for future study.

Development

The problem of describing man's walking performance under reduced gravity conditions is approached in three phases:

1. Description of normal walking
2. Selection of an experimental approach
3. Experimental evaluation of man's walking performance.

As an introduction to the three phases of this study, Chapter II is devoted to a review of previous walking studies. The more significant findings of previous researchers are noted along with a description of the techniques by which they attained their results. A more complete description of the merits and objectives of other experimental methods completes the chapter.

In accordance with the first phase of this study, a description of normal walking is presented in Chapter III. This description was deemed necessary to define terminology, to establish the characteristics of the gait that are significant to this study, and to provide a basis for comparison of the low-gravity gait.

Based on the available experimental techniques listed in Chapter II and the significant characteristics of the gait that are presented in Chapter III, a force measuring walkway is proposed in Chapter IV as the best means for measuring the low-gravity walking performance. However, delays in material procurement precluded its use in the present study.

A motion analysis technique was conceived as an alternate approach. Evaluation of low-gravity walking performance by means of a time-displacement walking experiment is presented in Chapter V.

Finally, some concluding remarks about the objectives of the study and recommendations for future work are given in Chapter VI.

II. Previous Studies of Walking

Normal Walking Studies

Although walking behavior has probably been studied for centuries, the first significant information dates back approximately one hundred years. The scope of these studies has ranged from a dynamics problem in classical mechanics to the correction of pathological walking behavior in the field of medicine. Certain of these works are generally accepted as milestones in the historical account of walking studies, and will therefore be mentioned here.

In 1836 the Weber brothers in Germany reported on anatomical studies on cadavers and on living subjects. In addition to determining the vertical center of gravity of the body, they presented a theory of walking and running in which they concluded that the swinging phase of the gait is a pendulum motion only, and does not depend upon muscular action. This theory gave rise to much discussion by other investigators, but was later repudiated by Braune and Fischer, Marey, and others (Ref 6:A-3).

Muybridge, in 1882, made photographic studies of the paces of the horse. Although one purpose of the study was to decide a bet as to whether all four legs were ever off the ground simultaneously, the study is significant because it was probably the first time that photography was used in the study of locomotion (Ref 6:A-6).

Between 1873 and 1895 in France, Marey's studies led to two important investigative techniques. The first innovation was the use of pneumatic cells (or "tambours") for the objective recording of forces and motion. Marey's greatest contribution, however, was the development of chronophotography - a technique by which successive exposures are made on the same photographic plate by means of a rotating mechanism within the camera (Ref 6:A-6).

Braune and Fischer, in 1890, determined masses and centers of gravity of the body and its various segments by measuring and dismembering cadavers. They established the first table of coefficients that related the dimensions and masses of the body segments to overall height and weight (Ref 6:A-8).

Later, between 1898 and 1904, Fischer published six volumes concerning the human gait. By use of four chronophotographic cameras simultaneously to photograph miniature gas tubes located at specific body points, he was able to record the displacements of the body during 31 phases of the double step. The results of his detailed motion analysis indicated that the leg swing is not purely a pendulum motion, but that it also depends on muscle action. "Fischer's Der Gang des Menschen (Human Gait) is considered the classical work on gait" (Ref 6:A-10).

Between 1928 and 1936, Bernstein in Moscow repeated and expanded Fischer's experiments to include additions of mass to the body and also to study the effects of fatigue on locomotion (Ref 6:A-17).

Steindler, in 1935, published a comprehensive work on the mechanics of locomotion, primarily for use as a basis for the analysis and comparison of pathological walking gaits. The description of normal walking contained in this book (Ref 19:348-386) was used extensively as a reference during the early stages of this investigation.

Since 1945, much work has been sponsored by the National Research Council directed toward the improvement of the design and operation of artificial limbs. Although several universities (Refs 6, 8) have participated in this project, researchers at the University of California Biomechanics Institute (formerly the Prosthetics Devices Research Project) have published the most comprehensive study of normal walking (Refs 1, 3, 6, 14, and others). These reports were referenced extensively during the later stages of this investigation.

Low-Gravity Walking Studies

Personnel of the Crew Stations Branch of the Aerospace Medical Research Laboratories, Wright-Patterson AFB, Ohio, have conducted two zero-gravity walking programs and several low-gravity walking experiments aboard aircraft that were flown through Keplerian trajectories to produce periods of weightlessness. Because man can not walk unaided at zero-gravity, special footwear was designed for, and was the subject of, the zero-gravity tests.

A study of magnetic shoes was conducted during 1959. Although the magnets provided sufficient vertical force to hold the man to the surface, walking was impossible because the metal-to-metal friction of the magnetic sole against the surface was insufficient to permit the subject to develop the necessary fore-and-aft shear propulsive forces. Several improved designs of magnetic shoes have since been proposed, but have not been tested under weightless conditions (Ref 18).

A subsequent test was made using Velcro material on the shoe soles and on the walking surface (Ref 17). Although the Velcro shoes provided the necessary holding and shear forces, walking was possible but still difficult. The absence of the gravity acceleration to aid body and limb rotation requires that some muscles must perform functions for which they are ill-fitted. For example, forward body rotation must be accomplished by the pre-tibial muscle group which must reverse its normal restraining action and must instead pull the shank of the leg forward about the ankle.

Although these tests provided information applicable to the use of Velcro and magnetic devices in other space applications, the necessity for special walkway surfaces and shoes in space applications degrades their practicality as an aid to walking. In a weightless state, locomotion near a surface will probably be performed, more easily, either by soaring or else by pulling one's self forward hand-over-hand along a series of recessed handholds.

An additional low-gravity walking experiment was performed during 1961 to determine the lower gravity limit at which man can still walk. Based upon the observation of only one subject, the lower limit for acceptable walking was determined to be in the neighborhood of 0.2 g (Ref 10:19). Later tests have shown that man can still walk at lunar gravity (0.17 g). However, all these experiments have been qualitative in nature, and have provided no quantitative description of walking performance at these gravity levels.

Experimental Methods and Instrumentation

Knowledge of the relationship between the forces acting on the body and the resulting body movements is fundamental to the understanding of locomotion. Instrumentation techniques that have been used to obtain this information can be divided into two groups: instrumentation for motion analysis, and instrumentation for force analysis.

In addition, mention will be made of some previously used walkways.

Instrumentation for Motion Analysis. Most motion studies have been recorded by photographic techniques ranging from still photography to high speed motion picture photography. Motion pictures provide an excellent means of recording motion. By simultaneously photographing a timing device (or by counting frames) the recorded displacements can be plotted as a function of time and the velocity and acceleration of selected points can be determined.

Often some form of photographic targets, such as black dots on white tape, are fixed on the skin at body landmarks to facilitate data reduction. In some studies of leg segment transverse rotations, long stainless steel pins were imbedded into the various bones to mechanically amplify the small rotational displacements (Ref 9:859) (see "Body Axis System," Chapter III, for an explanation of the body axes, orientation planes, and transverse rotations).

An improved method of chronophotography called the method of interrupted lights has been found useful for obtaining "stick" diagrams of the walking displacements. Small ophthalmic lights, powered by a hand-carried battery, were attached to the subject at joint locations. The subject walked in a darkened room in front of the camera whose view was interrupted 30 times a second by means of a rotating disc.

The resulting photograph showed the location of the selected points at specific time intervals, and made possible an accurate determination of velocity and acceleration (Ref 6:Chapter 1).

Linear and angular accelerometers have been attached to the subjects in some motion studies to provide a means of checking the accuracy of the photographic reduction process. It was found that the accelerometer data required more reduction time and was less accurate than the data obtained by photographic reduction (Ref 3:1127).

Information on joint angles during motion has been most easily obtained by the use of electrogons (Ref 5, 7). The electrogon is an electrical potentiometer that indicates the relative angle between two adjacent segments when it is strapped across a joint. The oscillograph recording is called a goniogram. Although electrogons have been used primarily for recording knee and ankle movements during walking, some athletic motions were also measured. For example, the elbow angle at the instant of baseball release by a professional pitcher was determined to be 102° - much less than the presumed 180° (Ref 7:10).

<u>Instrumentation for Force Analysis</u>. The direct measurement of walking forces requires a data sensor (or transducer) and a recorder. An oscillograph is generally used to record the electrical signals from the transducer.

Small resistance and capacitance force discs have been placed on the soles of the feet to record pressure patterns during walking. Because the foot contact area and spatial orientation are continually changing, the oscillograph tracings do not indicate the total force on the foot, but only indicate the degree of weight bearing by each disc. No attempt has been made to obtain quantitative force information by this method. The podograms (or "electrical foot-prints") obtained were used chiefly to compare normal and abnormal gaits (Ref 16:203-214).

Force plates have been used in several investigations for the direct measurement of the surface reaction forces (Refs 6, 13, 16). The force plates usually consist of a surface deck which is supported by strain-gaged cantilever beams or columns. Proper design of the beams or columns permits accurate measurements to be made of the forces and moments about any of three axes simultaneously. Oscillograph tracings at a known paper speed provide force versus time data.

Strain-gaged pylons have been inserted as the structural member in the shanks of artificial legs (Ref 6:Chapter 7). The pylon method has the unique advantage that it measures leg forces and moments during the swing as well as during stance. However, its application is limited to amputees, and will not be further considered here.

Phasic muscle activity has been successfully recorded by the use of electromyographic techniques. The action potentials of the muscles are picked up by surface electrodes, are amplified by a modified electroencephalograph, and are recorded by an oscillograph. Although this technique gives only qualitative information concerning the forces generated within the muscles, the period during which the muscle is acting is precisely determined.

Walkways. Mention should be made of some of the previously used walking surfaces. Some walkways consisted of little more than the floor or an elevated platform (Refs 5:iii, 9:860). A distance scale (spaced pegs or floor markings) was usually included to facilitate photographic reduction. In addition to level walking surfaces, work and energy studies have often been performed by having the subject ascend or descend a ramp or stairs.

Treadmills have been used in many laboratory tests of walking (Ref 5:iii). They are particularly useful when it is desired to control the walking velocity and/or sustain the walk over a considerable duration. The proximity of a heavily instrumented subject to the recording equipment is often a distinct advantage, especially during evaluations of body metabolism during walking.

Certain of the above techniques have been used concurrently during some tests. Perhaps the most comprehensive results were those obtained by researchers at the University of California, who made a three-dimensional motion picture study of subjects walking on a long glass walkway into which was imbedded two force plates (Ref 6:Chapter 8).

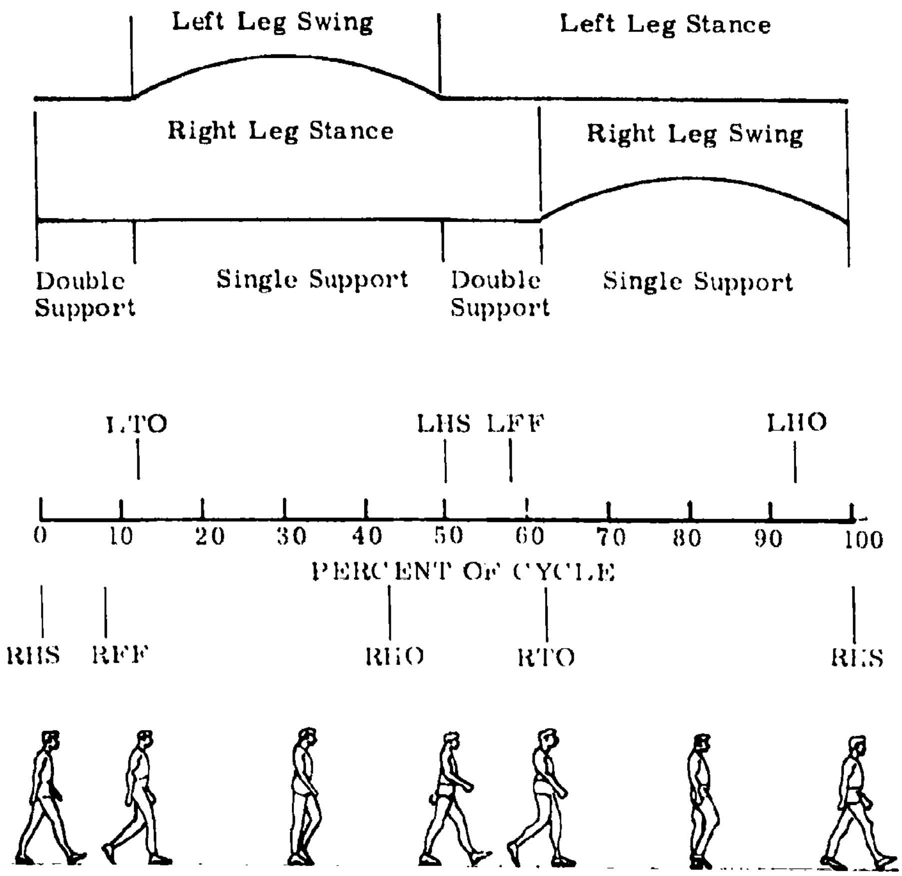
III. Description of Normal Level Walking

The human walk[1] consists of alternate periods of support and swinging motion by each leg, with the end effect of progressive translatory motion of the body. The bipedalism peculiar to man requires that the legs alone must assume the functions of support and propulsion, as follows: in alternating play one extremity is placed to the ground in an oblique direction in a flexed position. Then, by sudden extension, it imparts to the body center of mass a propulsion forward and upward. At the same time, the other leg swings forward so that, in order to meet the downward tendency of the body path, it may effect a restraining action at the moment when it has finished its swing and is again set to the ground (Ref 19:355). The legs alternate in these events in such a way that the supporting phase of one leg largely coincides with the swinging phase of the other.

Phases of the Walking Cycle

The time that a foot is in contact with the ground is called a support phase. This is shown in Fig. 1. The support phase begins

[1]Walking, as used in this report, is contrasted from running, jumping, sprinting, or soaring, by the requirement that at least one foot must be in contact with the surface at all times.



LHO - Left Heel Off
LFF - Left Foot Flat
LTO - Left Toe Off
LHS - Left Heel Strike

RHO - Right Heel Off
RFF - Right Foot Flat
RTO - Right Toe Off
RHS - Right Heel Strike

FIG. 1 PHASES IN THE WALKING CYCLE

at the moment the heel is set to the ground and ends with the "takeoff," i.e., the moment the big toe leaves the ground after the deploy of the foot. The swing phase begins when the toe leaves the ground to swing forward, and it ends at the moment when the heel again strikes the ground. The period during which one limb accomplishes the above two phases is called a double step. The distance covered during this period is called the stride.

In walking, the time of support is longer than the swinging time. Finley and Karpovich determined that the ratio of swinging time to support time varied from 0.5 to 0.8 for normal subjects (Ref 5:7). Because the swing-to-support ratio is less than one, there are transitional periods when both feet are in contact with the surface. These are known as periods of double support. At average walking speeds, the ratio of the period of double support to the period of single support is about 4:10 (Ref 19:358). (This corresponds to a swing-to-support ratio of 0.55.)

The swinging leg is often compared to a pendulum. As stated in Chapter II, the Weber Brothers concluded that the swinging phase of the gait was a "pure" pendulum motion and did not depend upon muscular action. By later experiments and analysis, Marey, Fischer, and others repudiated this theory and acknowledged the presence of other factors. The accuracy of the approximation is improved by further restriction; for example, the leg may be described as a compound physical pendulum. The analysis on this basis, however, is complicated by the facts that the leg is attached to a moving support, the distribution of its mass segments and their respective distances from the moving support changes as the leg is shortened by joint flexion during mid-swing, and the internal moments that are created by the muscle forces are both nonlinear and time varying. Knowledge of the magnitudes and the time variation of all these factors is essential for the computation of the motion of the swinging leg. Thus, the swinging leg must only be considered as a unique physical system which is acted upon by the various forces and moments and is constrained to definite boundary conditions and the path of the moving support.

Despite the restrictions that are necessary for a rigid mathematical analysis, the motion of the swinging leg is most easily visualized as that of a pendulum. Therefore, this analogy is frequently used in the qualitative description of walking.

Notwithstanding the above restrictions, the swinging leg, like a pendulum, tends to have a natural frequency and a period that is dependent upon gravity and the length of the leg. Regardless of how fast a man walks, the swinging time of his leg remains essentially constant. During a fast walk the amplitude increases, but the period remains the same. As walking speed increases, the time of support decreases. But, consistent with the definition of walking, the swinging time of the leg can never exceed the support time (Ref 19:356).

The propulsive force that is necessary to keep this motion going is introduced by the foot as it pushes backward against the surface just before it deploys into the swing phase. A decelerating force is introduced when the heel strikes the ground at the end of the swing to become the new point of support.

Body Axis System. A three-coordinate, right-handed, orthogonal body axis system is defined with its origin at the center of mass of the standing man (see Fig. 2). In this system, the positive X axis is directed forward, the positive Y axis is directed toward the left, and positive Z axis is directed vertically upward. These three axes are the intersections of the three cardinal planes of the body. The cardinal transverse plane is a horizontal plane which contains the X and Y axes and divides the body into upper and lower halves. The cardinal frontal plane contains the Y and Z axes and divides the body fore and aft. The cardinal sagittal plane, which is usually termed the "plane of progression" in discussions of locomotion, contains the X and Z axes and divides the body into left and right halves.

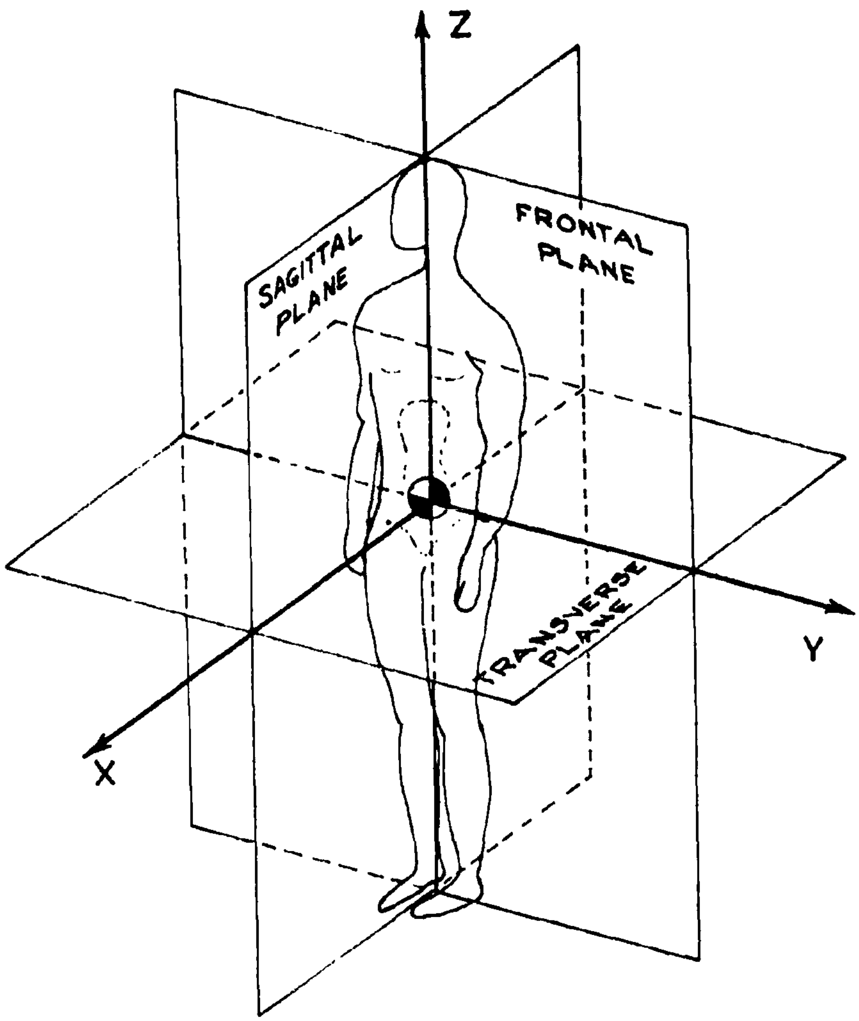


FIG. 2 BODY AXIS SYSTEM AND CARDINAL PLANES

In addition to the body axes and cardinal planes, local axes and planes that are parallel to the body axes and cardinal planes may be defined with their origins at any desired body location. Thus, a "transverse rotation of the ankle" is a rotation of the ankle about the Z axis of a local coordinate system which has its origin at the ankle joint. This movement would occur in the local transverse plane of the ankle.

Forces in Normal Level Walking

Man, in walking, may be considered to be an isolated body acted upon by the external forces of weight (due to the acceleration of gravity acting on the body mass), air resistance, and the reaction forces of the surface against his feet (see Fig. 3). Internal forces will not be considered in this analysis except to acknowledge that they provide for body balance and the changes in position of the limbs necessary for walking.

External Forces.

Weight. At first glance the weight force, because it is directed perpendicularly to the line of progression, would seem to have little effect on normal walking. However, further analysis quickly emphasizes its significance. The force due to gravity applies uniformly over the human body. This force acts on any segment of

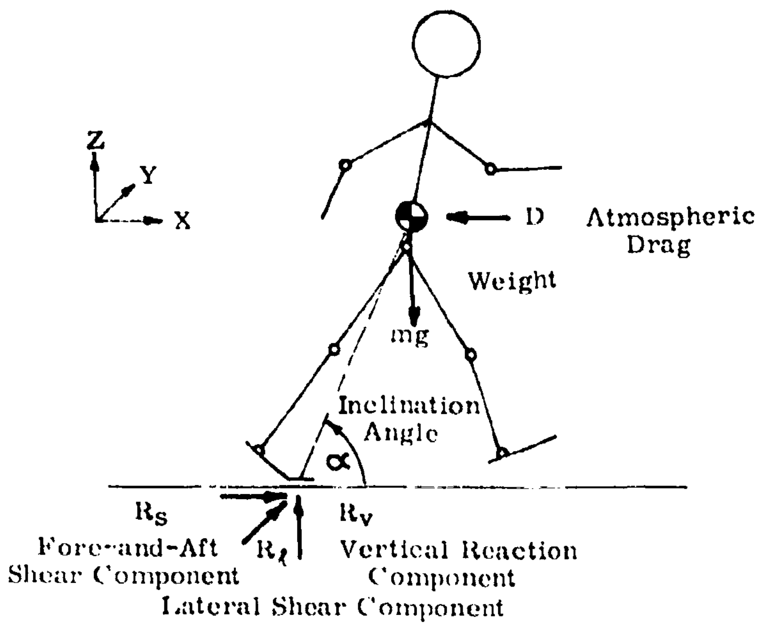


FIG. 3 EXTERNAL FORCES ON THE WALKING MAN

the body to produce a moment about its respective joint axis of

$$M_g = m\ g\ k \cos\alpha \qquad (0)$$

where

M_g is the gravity moment,

g is the local gravity acceleration,

k is the distance between the segment center of mass and the respective joint axis, and

α is the inclination angle of the segment longitudinal axis measured from the horizontal.

By assuming an inclination of the various segments, or indeed, of the body as a whole, the weight force becomes a driving force which tends to accelerate the body in angular rotation. For example, when the swinging leg is ahead of the stance leg the weight force tends to rotate the body forward about the point of support, resulting in forward translation of the body center of mass. The gravity force is also considered to be the primary force that drives the swinging leg through its characteristic "pendulum-like" motion. The combination of the weight force and the forward inclination of the body opposes the external resistance developed by atmospheric drag on the body. The resultant of the surface reaction forces during the "push-off" phase is directed below the center of mass of the body, and tends to rotate the body backward. Therefore, the combination of the weight force and the forward inclination of the body also acts to prevent the trunk from falling backward when the impetus of the "push-off" leg is transmitted to the body.

In addition to the above driving and restraining functions, one must not overlook the fact that the gravitational force "holds" man to the walking surface. Without the combination of this holding force and friction, man would be unable to anchor his foot to develop the propulsive fore-and-aft shear forces.

Atmospheric Drag. All movement through an atmosphere is opposed by atmospheric drag. During walking, the body opposes this drag force by assuming a forward inclination. Uniform progressive motion, according to Newton's first law, requires no external force. Many investigators have proposed that air resistance is the reason that man must continually apply impulses at his feet to maintain his walking velocity. However, consideration of the magnitude of the drag force does not justify this conclusion.

The atmospheric drag on a man walking under sea-level conditions at a velocity of 5 fps is computed to be 0.71 pounds (see Appendix A). The forward inclination of the body that is necessary to balance this drag moment is only 0.25 degrees. This corresponds to a shift of the body center of mass 0.18 inches forward of the "no-wind" equilibrium position. It is considered by the author, therefore, that the role of air resistance at normal walking speeds has been over-emphasized, and that it is negligible in comparison to other factors. (In any case, it is below the accuracy of the proposed force-measuring systems to be used in this study.)

Surface Reaction Forces. The walking surface must support the man and provide sufficient traction to enable him to develop a propulsive force. Thus, there is at all times in walking, a surface reaction force which is equal and opposite to the vector sum of the forces of weight, air resistance, and the inertial forces due to the acceleration of the body mass. This surface reaction force is resolved into a vertical component (R_V) that opposes weight and the up-and-down inertia forces, a lateral shear component (R_ℓ) that opposes the side-to-side inertia forces, and a fore-and-aft shear component (R_S) that opposes air resistance and the fore-and-aft inertia forces of the body mass. The vertical component is defined as positive when it acts in the positive Z direction (upward). The lateral shear component is defined as positive when it acts in the Y direction (left). The fore-and-aft shear component is defined as positive when it acts to prevent the foot from sliding backward, i.e., when the shear force tends to accelerate the body in the positive X direction (forward).

Typical magnitudes of the surface reaction forces during a complete walking cycle are graphed in Fig. 4 (Ref 6:Fig. 8-33). These curves represent the forces that act during the "steady state" portion of the walk. The curves for accelerative walking would probably have the same shape but different magnitudes. The forces are plotted on the ordinate as a function of the percentage of cycle as the abscissa. The cycle may be considered as "normalized" time. Walking at a step frequency of 95 steps per minute, the full cycle requires 1.26 seconds of time. The solid lines are drawn for the right foot, whereas the dotted lines denote the left foot.

The vertical force component has a double-peaked shape because of the vertical upward and downward accelerations of the body. The difference between the magnitude of the vertical force and the body weight is proportional to the vertical acceleration. As the heel strikes the ground the force rises quickly to a magnitude equal to or slightly greater than the body weight. The unlocking of the stance knee results in the dip in the curve. As the trunk of the body moves vertically over the stance leg, the knee flexes to prevent the body from vaulting. At the same time, the hip on the side of the swinging

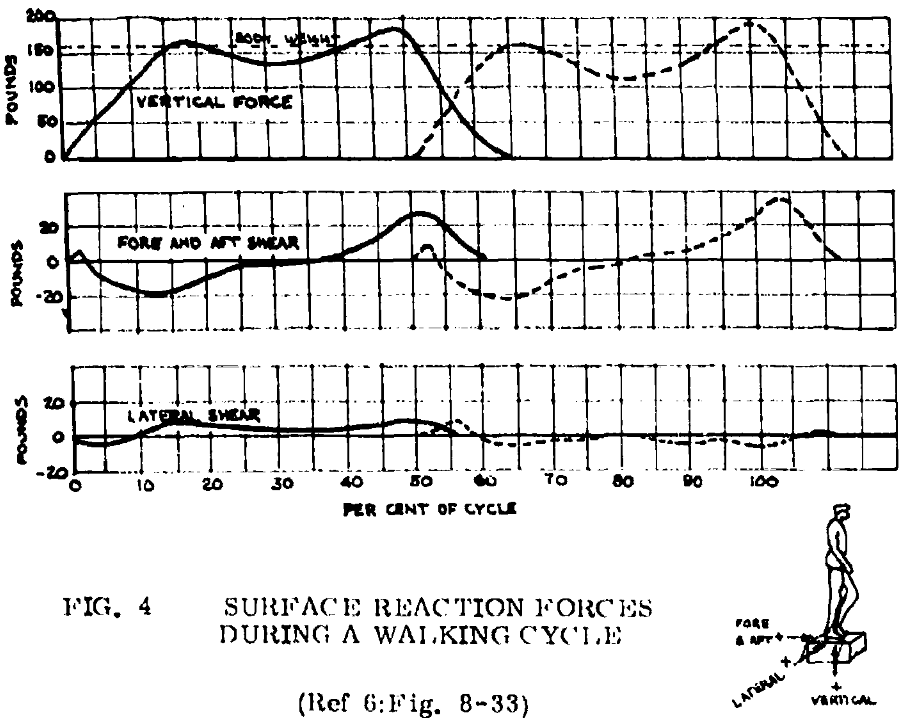


FIG. 4 SURFACE REACTION FORCES DURING A WALKING CYCLE

(Ref 6:Fig. 8-33)

leg is lowered. The combination of these events enables the subject to move forward with a minimum raising of his center of mass, and therefore requires less work. As the leg prepares for the push-off phase, the knee again locks. Extension of the knee and ankle at this point imparts the forward and upward acceleration to the body. This causes the curve to again exceed the body weight.

The fore-and-aft shear curve has a sinusoidal shape. The variation in the force is due to the fact that upon heel strike the leg must first retard the forward motion of the body, and then a fraction of a second later, must provide the "push-off" or forward acceleration necessary to continue the motion. The small positive "blip" at the start of the curve implies that near the end of the swing the leg has decelerated to zero and the foot is actually moving aft at the instant of heel strike.

The lateral shear force is of relatively small magnitude but is important to provide lateral stability in walking. As the body weight is shifted from one leg to the other, the lateral force must be directed inward with respect to the body to prevent the subject from falling sideways. The force is thus positive for the right foot, and negative for the left foot, respectively, at all times after the foot is flat on the ground.

Internal Forces. Although the internal forces of the body need not be considered in a motion analysis of the body considered as a free body, some discussion will greatly facilitate the reader's understanding of the human walk. The propelling forces for walking must be developed internally. The muscles cannot directly translate the whole body because they have no point of attachment outside of the body. The muscles do, however, exert moments about the joints which may act either to prevent or to cause relative rotation between the segments on either side of the joint. The combination of rotations of all the various body segments produces the net result of forward motion (Ref 1:3).

Muscle Groups. The muscles that are of primary interest in the study of walking are those of the lower extremity. These muscles have been classified by researchers at the University of California into the following eight groups:

1) Pre-Tibial group,
2) Calf group,
3) Quadriceps group,
4) Hamstring group,
5) Abductor group,
6) Adductor group,
7) Gluteus Maximus group, and
8) Erector Spinae group.

The anatomical locations of these muscle groups are shown in Fig. 5 and their respective functions are schematically diagramed in Fig. 6 and Fig. 7.

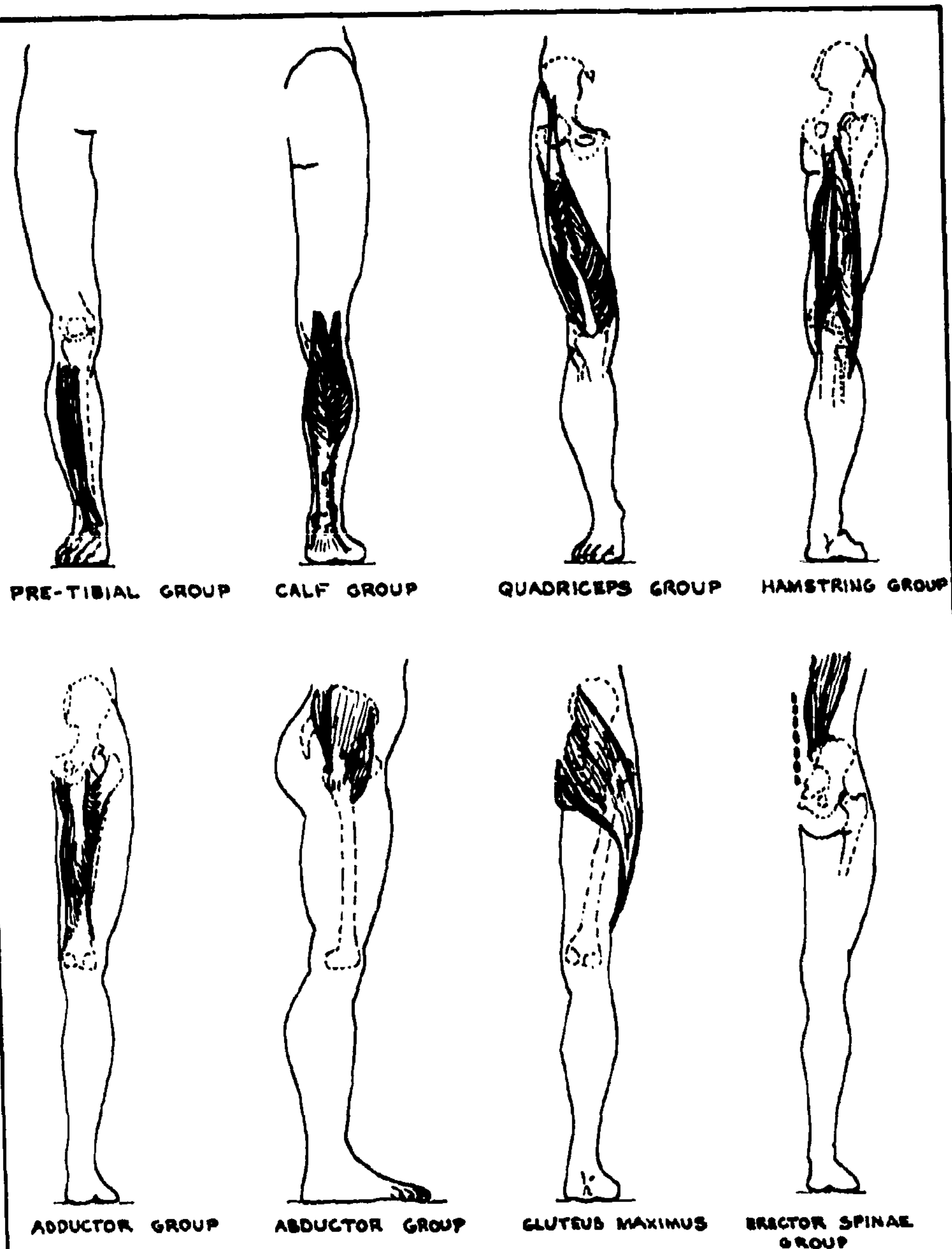


FIG. 5 MUSCLE GROUPS

(REF 6 : FIG. 3-5)

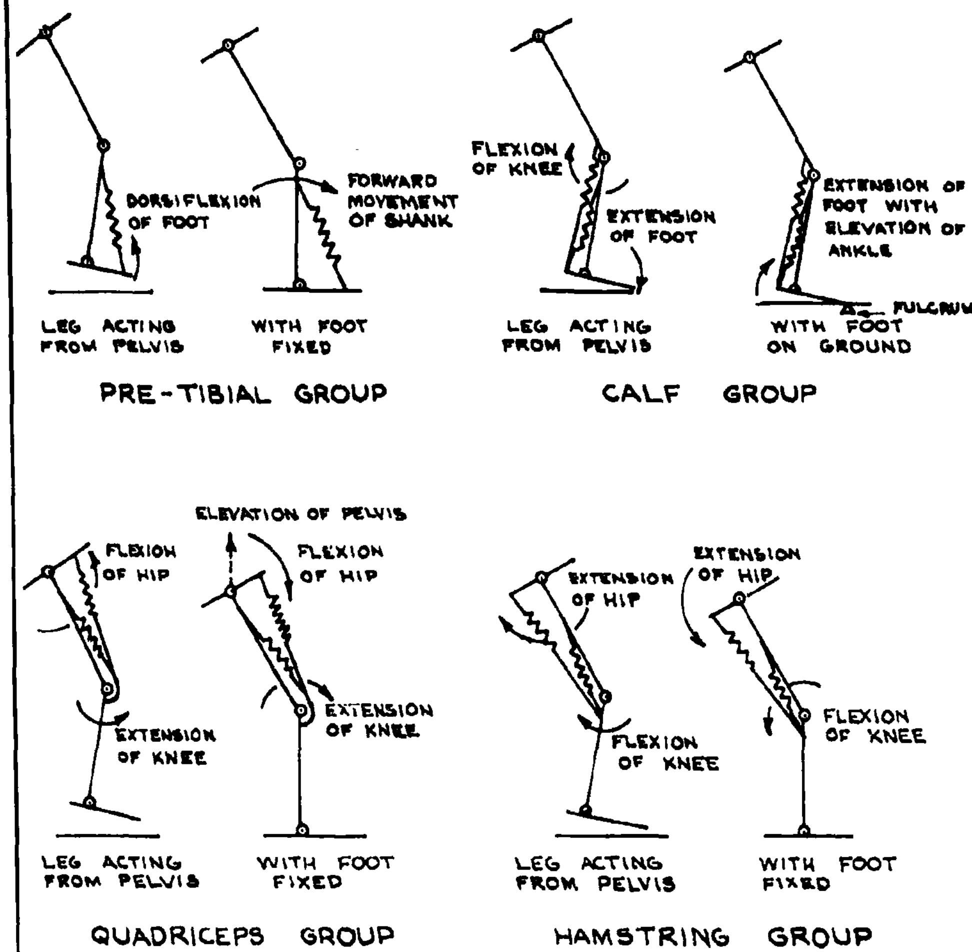


FIG. 6 SCHEMATIC MUSCLE ACTION DIAGRAMS - I

(REF 6 : FIG. 3-6)

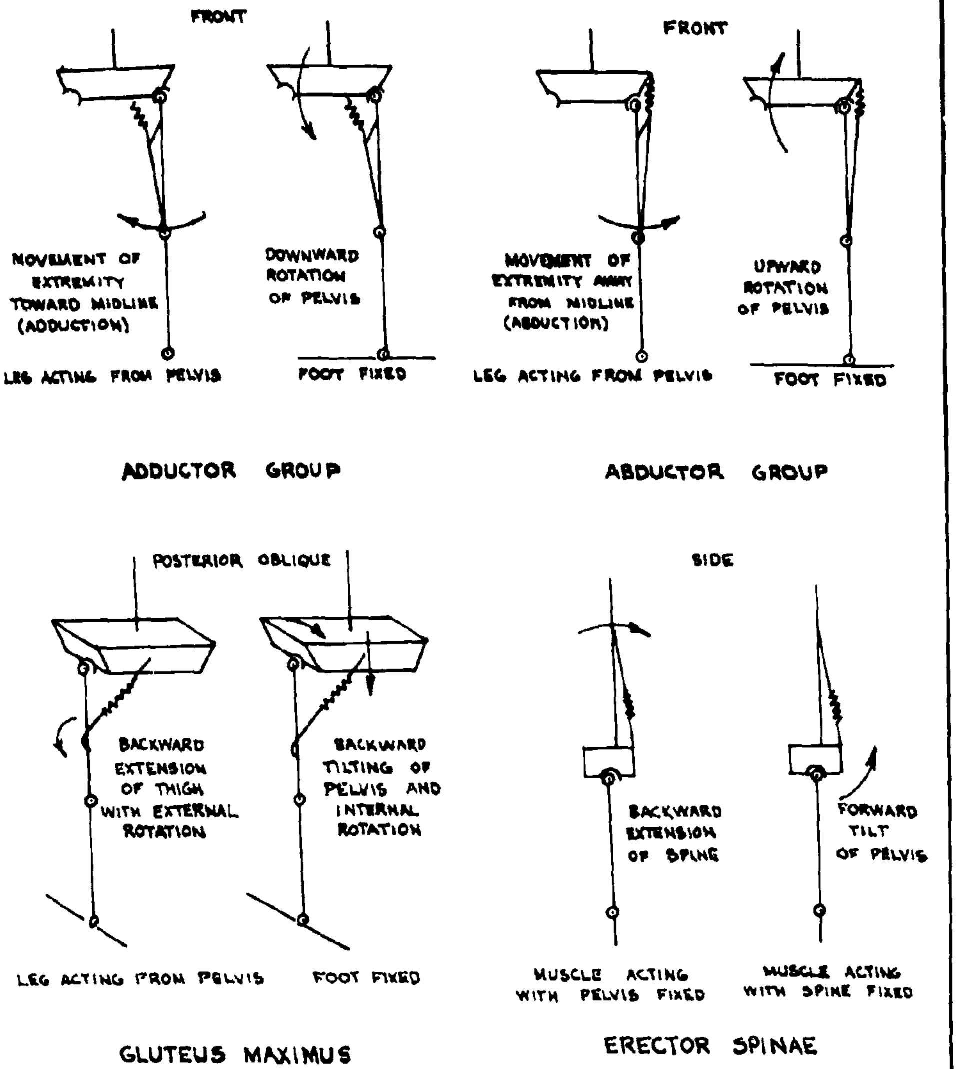


FIG. 7 SCHEMATIC MUSCLE ACTION DIAGRAMS-II

(REF 6 : FIG. 3-7)

The phasic action of the eight muscle groups was analyzed by use of electromyographic techniques. Electromyography reveals the precise phase relationship of muscle action. However, it gives only suggestive or qualitative information concerning the forces generated within the muscles. The action potentials of the muscles are picked up by skin electrodes, are amplified by a modified electro-encephalograph, and are recorded by an oscillograph. Thus, a record is obtained showing which muscles are acting during each phase of the walking cycle.

Muscle Functions. The functions of the major muscle groups were qualitatively deduced from the electromyograph tracings to be:

1) Pre-Tibial Group: The maximum activity of the pre-tibial group occurs very shortly after heel strike when the muscles act to lower the foot to the surface without slapping. Despite their contraction at heel strike, the muscles are being lengthened as they resist the moment caused by the body weight acting about the fulcrum at the heel. The resistance of the pre-tibial group initiates the deceleration of the leg and body following heel strike. (The quadriceps group completes this deceleration after the foot is flat on the surface.) The secondary function of the pre-tibial group is to dorsiflex the foot at toe-off to

provide maximum clearance between the toe and the walking surface during the swing phase. This action also rotates the foot to the optimum angle for the impending heel strike (Ref 6:3-19).

2) Calf Group: The calf group muscles act only during the stance phase and perform three functions. First, as the body weight passes forward of the ankle after mid-stance, the calf group resists ankle flexion and thereby causes the heel to come off the floor. This moves the center of pressure of the floor reaction forward from the ankle to the ball of the foot. The second and most significant function of the calf group is to extend the foot during the push-off phase, propelling first the entire body mass and then later the swinging leg. The third function of the calf group is to flex the knee of the stance leg to control the vertical height of the body as it rotates over the stance foot (Ref 6:3-23).

3) Quadriceps Group: The quadriceps group acts mainly during the stance phase. As the stance foot reaches the foot-flat position, the quadriceps group takes over control of the deceleration of the downward movement of the center of mass by permitting controlled flexion of the knee. The quadriceps muscles also contribute to the push-off force. By extension of the knee with the foot fixed, they accelerate the body in an upward and forward direction. At

faster walking speeds, the quadriceps act to retard and reverse the backward oscillation of the leg at the hip, resulting in forward acceleration of the leg for its swing-through activity (Ref 6:3-26).

4) Hamstring Group: During normal walking, the hamstring group acts during the swing phase to terminate the forward velocity of the swinging leg and to plant the heel to the surface. As the decelerative force of heel strike is transmitted up the leg, the hamstrings act to oppose the gravitational and inertial moments of the upper torso that are tending to flex the hip forward. At faster walking speeds, the hamstrings act during the swing phase to shorten the swinging leg by knee flexion (Ref 6:3-29).

5) Abductor Group: (Abduction is the act of moving an extremity laterally away from the midline of the body.) The abductor group acts during the stance phase to prevent the downward list of the swinging-leg side of the pelvis as the stance leg assumes the full weight of the body. The abductor group also limits the side-to-side oscillation of the body center of mass by opposing and limiting adduction (Ref 6:3-30).

6) Adductor Group: (Adduction is the act of moving an extremity laterally toward the midline of the body.) The adductor group acts primarily during the early stance phase. During the

period of double support, the combination of the forward leg in the heel-strike phase and the aft leg in the push-off phase develops a "scissors" torque on the pelvis. The adductor muscles resist this torque and thereby minimize the horizontal rotation of the hips. During late stance, the adductors rotate the femur outward (toe out) in preparation for the swing. Because of their location across the hip joint, the adductors aid other muscle groups in controlling both extension and flexion of the hip. The adductor muscles, in combination with gravitational forces, also control inward motion of the leg and hip in the frontal plane (Ref 6:3-33).

7) Gluteus Maximus: The gluteus maximus is a hip extensor muscle that may act at any time it is needed to resist forward rotation of the hip. Its primary activity occurs at heel strike when, like the hamstrings, it prevents the hips from rotating forward. It also resists the horizontal rotation of the pelvis about the stance leg during the push-off phase (Ref 6:3-35).

8) Erector Spinae Group: The erector spinae group acts throughout the entire walking cycle, but its peak activity occurs during heel strike of either foot. At this time it resists the gravitational and inertial moments that tend to rotate the upper torso forward about the pelvis. In addition, the erector spinae also balances the upper torso on the pelvis as the body sways from side to side in the frontal plane (Ref 6:3-35).

Displacements During the Walking Cycle

The displacements of the body during the walking cycle are the result of the complex motion of the parts of the body in three planes. A complete description of the displacements of the various segments of the body is beyond the scope of this report. However, an examination of the displacement of the body's center of mass will provide a basis from which to investigate other specific characteristics of the walking gait.

In level walking, the center of mass of the body describes a smooth regular sinusoidal curve in the plane of progression. The curve peaks twice during each cycle as the body passes successively in a double step over first the right leg and then the left. The total vertical displacement in adult males is about one and eight-tenths inches; the high points occur at the middle of the swing phases (25 and 75 per cent of cycle) and the low points occur during the periods of double support (50 and 100 per cent of cycle) (Ref 15:545).

A person is slightly shorter when he is walking than when he is standing. That is, if a person were to walk through a tunnel of a height exactly equal to his standing height, he would not bump his head because he would have nearly one-half inch of clearance (Ref 15:545).

The center of mass is also displaced laterally in the transverse (horizontal) plane. Relative to the plane of progression, the center of mass describes a sinusoidal curve, the peaks of which alternately pass to the left and right in association with the support of the weight-bearing leg. The magnitude of the displacement in normal level walking is approximately one and three-quarters inches from peak to peak.

If the vertical and horizontal displacements of the center of mass were combined and projected onto the frontal plane, they would describe an almost perfect figure of eight, occupying approximately a two-inch square, because the vertical and horizontal deviations are almost equal (Ref 15:545).

IV. A Proposal for a Force Measuring Walkway

Introduction

Walking behavior may be studied by the use of either a force analysis or a motion analysis. Knowledge of the relationship between the forces acting and the resulting body movements is fundamental for a general understanding of locomotion. Each may be determined from the other by analytical methods if it is assumed that the physical characteristics of the segments of the body such as mass distribution, length, and the moments of inertia are sufficiently well-known. However, the labor involved and the inaccuracies due to the assumptions and excessive calculation usually preclude such a study (Refs 2:32, 3:1214). The choice of a particular method is dictated by such factors as the simplicity of recording, ease of reduction of the data, and the accuracy of the results.

Previous studies of low-gravity walking have established that the lower limit of gravity at which man can still walk unaided is approximately two-tenths normal gravity (Ref 10:19). A motion analysis will confirm the subject's poor performance, but it will fail to explain why this is so. The acceleration of gravity acts on the body mass to produce one of the major forces of locomotion. It is only logical, then, that an analysis of walking in which gravity is a controlled variable can be made most directly by use of a force method.

The primary problem in a force analysis program is to measure the magnitude and direction of the forces between the feet and the walkway surface. Two methods of instrumentation, force discs and force plates, have been used before in the analysis of walking behavior. Force discs have several disadvantages. Because a force disc measures pressure and not force, the resulting pressure must be multiplied by the area of the foot in contact with the walking surface to obtain the force. However, the foot contact area is continually changing during the walking phases. The second disadvantage of the force disc is that the direction of the resulting force is unknown.

The force plate, in contrast, has the advantage that the directions of the surface reaction forces are measured directly. A force plate can be used with any type of footwear, and its surface can be coated with a material of any desired coefficient of friction.

Structural Design

In accordance with these basic characteristics, the principles of a force plate were extended in scale toward the design of a force measuring walkway. The walkway consists of two parallel platforms, each one foot by twelve feet, that are mounted on cantilever beams which are strain gaged to measure independently for each foot the vertical and fore-and-aft shear surface reaction components of a walking subject. The length of the platform permits a continuous-time oscillograph recording of the respective forces during several consecutive walking steps. A sketch of the proposed walkway is shown in Fig. 8.

A complete description of the proposed walkway is beyond the scope of this chapter. However, certain design factors were deemed significant and will be stated to justify the proposed configuration. The basic requirements of the force-measuring walkway were that

1. It must support the man at gravity levels between zero-g and normal gravity.
2. It must measure independently for each foot the two components of the reaction forces in the plane of progression.
3. It must be of a size and weight compatible with the KC-135 airplane and the desired test objectives.

By use of structural metals, the requirement for support was found to be less stringent than the requirements for rigidity of the overall walkway and the sensitivity of the cantilever beams. The

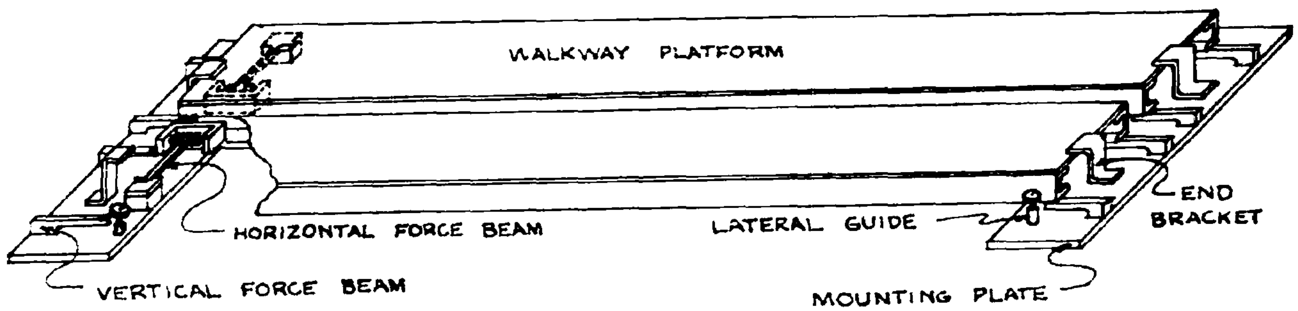


FIG. 8 FORCE-MEASURING WALKWAY

KC-135 floor-to-ceiling height is 78.5 inches. It was necessary to limit the height of the assembled walkway to approximately six inches to allow head clearance for the walking subject. To preclude a "springboard" action of the walk on the subject it was necessary to limit the maximum mid-span deflection of the walking platform to less than one-half inch. This deflection was selected arbitrarily as a gross approximation for the conditions that are stated in Appendix C, namely, that the subject stands at mid-span and is supported by only one of the two channels. The deflections and the natural frequency of the assembled platform should be further investigated.

The structural components of the force-measuring walkway consist of two walkway platforms, eight vertical force cantilever beams, two horizontal force cantilever beams, eight lateral guides, four end brackets, and two mounting plates. The more important design features of each component follow.

Walkway Platforms. Several construction materials were considered for the structural members of the walkway platforms. Steel was found to be too heavy, and wood could not support the load without excessive deflections. It was decided to construct the load-carrying members of the platform with standard aluminum channels, two per platform. Structural properties of 3-inch, 4-inch, and 5-inch channels in lengths between 8 feet and 15 feet were investigated. It was determined that a 12-foot length of 4-inch channel would limit the deflection to 0.375 inches for the critical condition of a 200-pound man standing at mid-span of only one beam (see Appendix C for sample calculations).

Commercially available 3/16-inch-thick aluminum abrasive tread plate was selected as the top plate, or walkway surface. The channels are spaced by 3/16-inch aluminum ribs at six-inch intervals along the length of the platform. Besides providing torsional rigidity and spacing, the ribs help support the walkway surface between the channels.

An assembled walkway platform is twelve feet long, one foot wide, and 4-3/16 inches high, and consists of two aluminum channels, an abrasive tread plate, and twenty-four ribs. Type 6061-T6 aluminum was specified for all platform components to permit fabrication by welding.

Cantilever Beams. The strain-gaged cantilever beams were designed to conform to accepted operating practice. Aluminum alloy is used instead of steel as the structural material because it has a greater stiffness for a given surface strain (Ref 12:242). That is, the aluminum beam will deflect less for a given load than a steel beam, provided that both are designed to yield the same maximum stress due to bending. Grade 75ST aluminum alloy was selected instead of 24ST alloy because its higher yield stress permits higher values of strain.

It is necessary that the beams retain definite end conditions. Therefore, anti-friction bearings were used at the free ends as an approximation of the theoretical knife edge (see Fig. 9). The fixed ends incorporate a drastic change in cross-sectional area between the point of anchorage and the calibrated portion of the beam to prevent "corner stresses" from being transmitted to the strain gages.

The beams were designed for a strain level of 2000 microinches per inch at the strain gage locations for a full-scale load. In the aluminum beams this corresponds to a stress (due to bending) of 20,000 pounds per square inch on the face of the beam at the locations of the strain gages. This value permits a high sensitivity, yet allows a sufficient margin of safety between the working stress and the yield stress of the alloy. This design condition resulted in a unit strain (strain per pound of load) of 8.5 microinches per inch for the vertical force cantilever beams, and a unit strain of 37.6 microinches per inch for the horizontal force beams. The deflection of the end of the

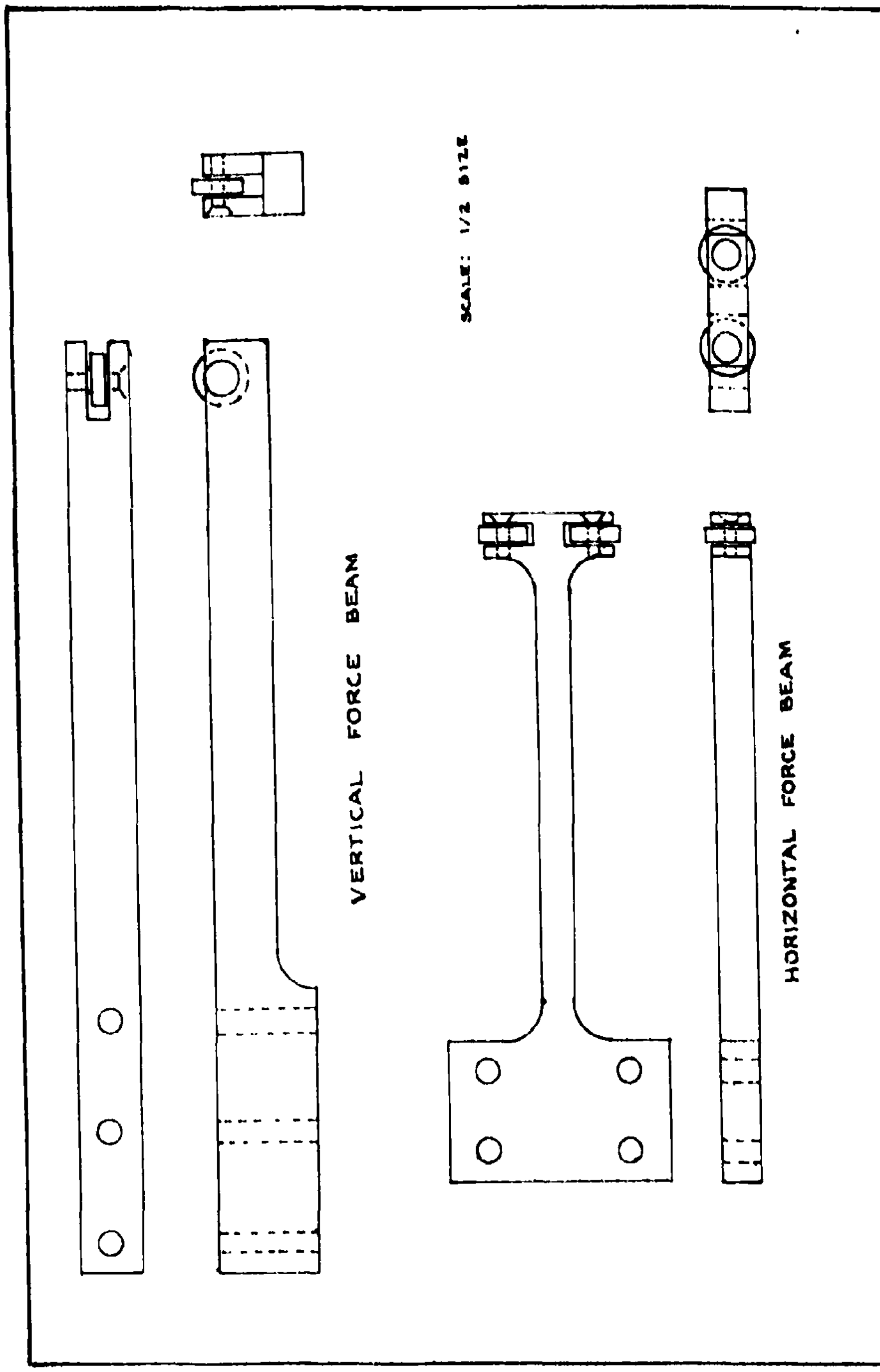


FIG. 9 CANTILEVER BEAMS

vertical force beam for the critical condition of the subject's weight being supported by only one beam was computed to be 0.073 inches. The deflection of the horizontal force beam due to the assumed maximum load of 50 pounds was 0.045 inches.

The vertical force cantilever beams support the corners of the walkway platforms and provide the means of sensing the vertical component of the walking forces. The horizontal force cantilever beam provides the means of sensing the fore-and-aft shear component of the walking force and also restricts the movement of the walkway platform in the X-direction.

Lateral Guides. The lateral guides restrict the lateral movement of the walkway platforms and yet permit small movements in the vertical and fore-and-aft direction, to allow for the minute deflections of the cantilever beams. This was accomplished by mounting self-aligning ball bearings in the horizontal plane with their rolling surfaces in contact with the sides of the walkway channels. The self-aligning bearings roll in the fore-and-aft direction and swivel in the vertical direction. Thus, the required two degrees of freedom are preserved with a minimum of friction.

End Brackets. The four end brackets are provided as a safety measure to prevent lofting of the platforms in case the airplane inadvertently enters a zero-g or a slightly negative-g maneuver. The end brackets should not contact the walkway platforms during normal ground tests or during normal flight maneuvers.

Mounting Plates. The two mounting plates function as a base to which all the other parts of the force-measuring walkway except the walkway platforms are attached. The mounting plates were designed to contain all the active components of the system to facilitate quick assembly of the system in the aircraft and to increase maintainability. Should circuit calibration or maintenance be required, it is only necessary that the mounting plates with their integrally attached components, and not the entire walkway system, be taken into the shop.

Electrical Instrumentation

The scope of the electrical instrumentation design is limited to the selection of the strain gages and the basic bridge circuits. The sensitivity of the circuits was computed as a basis from which to select the overall system components. The completion of the circuit, including the selection of the additional components and the mounting and calibration of the strain gages, is the responsibility of the Data Systems Division, Directorate of Test Data, Deputy for Flight Test, Wright-Patterson AFB, Ohio, whose personnel fabricate or approve all instrumentation that is placed in the KC-135 aircraft.

The electrical instrumentation for the force-measuring walkway consists of four independent measuring circuits; for each platform there is a vertical force circuit and a horizontal force (fore-and-aft shear) circuit. The outputs of these four circuits will be recorded on four channels of the 48-channel oscillograph that is permanently mounted in the airplane. Besides recording test data, the oscillograph is operated during all flights to record the accelerations of the aircraft about three axes.

Vertical Force Circuit. A vertical force-measuring circuit uses eight SR-4 strain gages. Two gages are mounted on each of the four vertical force cantilever beams; one gage is mounted on the upper face, and the other gage is mounted on the lower face. The eight gages are connected to form a four-active-arm, parallel Wheatstone bridge circuit. That is, two gages are connected in parallel in each arm. This arrangement gives an output that is proportional to the total vertical force on the platform, regardless of the location of the force. This is shown in Fig. 10. A resistance divider is included in the bridge circuit to balance the variations in the nominal gage resistances and their associated wiring.

The output voltage of the circuit for a unit vertical load on the platform was computed to be (see Appendix C for sample calculations)

$$E_o = 90 \text{ microvolts per pound.}$$

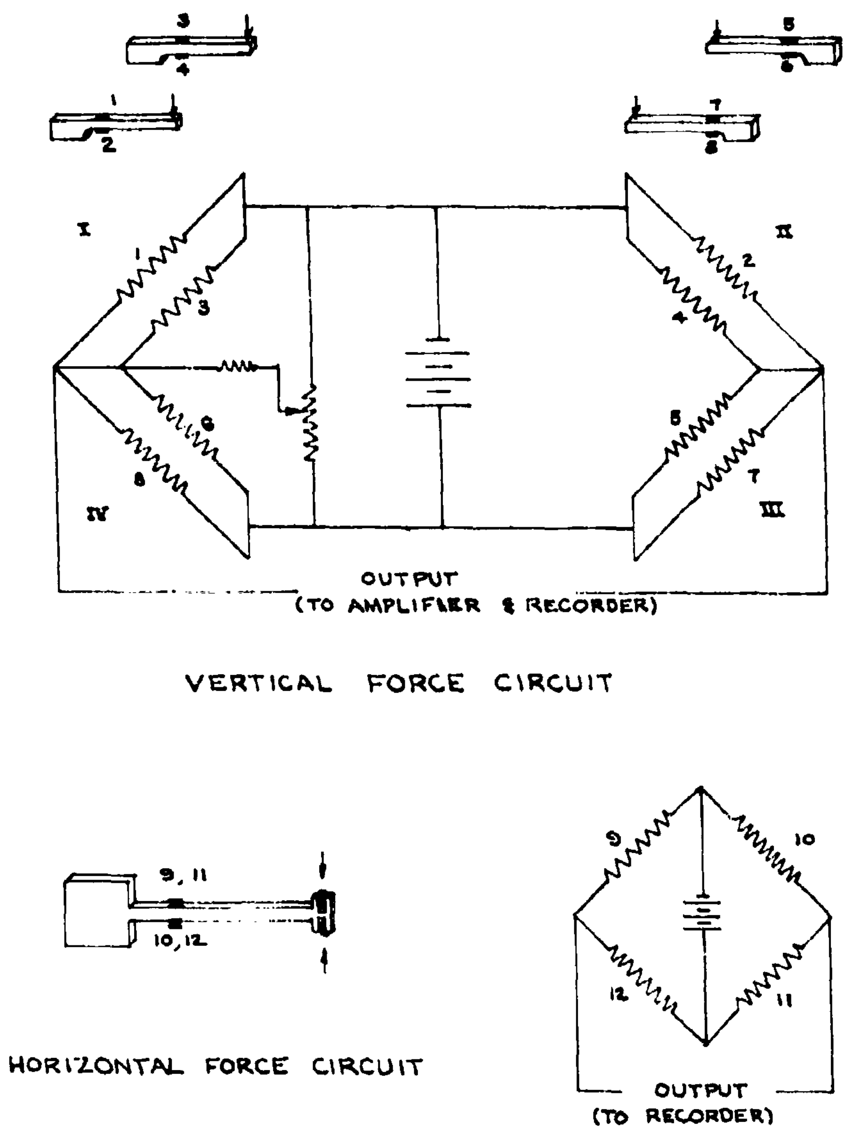


FIG. 10 STRAIN GAGE CIRCUITS

This low output will probably require amplification before it can be recorded.

Horizontal Force Circuit. The horizontal force-measuring circuit uses four SR-4 strain gages which are connected to form a four-active-arm Wheatstone bridge circuit. Two of the gages are mounted side by side on the forward face of the horizontal cantilever beam, and the other two are mounted on the aft face. It was decided to mount all four gages on the one beam to increase sensitivity and to prevent the occurrence of passive, or "dummy" branches.

The output voltage of the horizontal force-measuring circuit was computed to be

$$E_o = 1.58 \text{ millivolts per pound.}$$

This output is sufficiently great to permit direct recording.

Data Analysis

The velocity and acceleration of the walking subject may be obtained from the force-measuring walkway data by the use of Newton's second law

$$F = Ma \qquad (1)$$

which, when expressed as the principle of the conservation of linear momentum, becomes

$$F\Delta t = M\,\Delta V \qquad (2)$$

where

F = force, lb_f

M = mass, $lb_f\text{-}sec^2/ft$ (slug)

Δt = interval of time, sec

ΔV = change in velocity, ft/sec

$F\,\Delta t$ = impulse, $lb_f\text{-}sec$

The area under the fore-and-aft shear force curve may be obtained by graphical integration or by the use of planimeter techniques. The algebraic sum of this area at any time is equal to the net impulse applied to the walking subject during the same time interval. Therefore, the velocity of the man at any time may be found by dividing the net impulse by the mass of the subject (Ref 19:8-17). Differentiation of the velocity would yield the acceleration at any desired time. The accuracy of the velocity could be checked by performing a concurrent motion analysis (see Chapter V).

A diagram of the forces which act during one walking cycle is shown in Fig. 4 (Chapter III). The cycle was taken from the "steady-state" portion of the walk. Thus, the net area under the fore-and-aft shear force curve is zero, which correctly indicates that there was no change in velocity. No data are available concerning the accelerative and decelerative portions of the walk, that is, during starting and stopping.

Assume that a 161-lb man accelerates to a walking speed of 4 fps during his first step, which takes one second. The area under the fore-and-aft shear curve, by Eq 2, must be

$$F\ \Delta t = M\ \Delta V$$

$$= \frac{161\ (4)}{32.2} \tag{2}$$

$$= 20\ \mathrm{lb_f \cdot sec}$$

The reverse procedure is used in the reduction of the force-measuring walkway data.

The magnitude of the fore-and-aft shear force is limited by the product of the vertical force and the coefficient of friction. That is

$$R_s \leq \mu R_v \tag{3}$$

where

μ = coefficient of friction between the man's feet and the walking surface

and it is understood that the magnitude of R_v is dependent upon the gravity level. If the gravity level is reduced, the magnitude of the vertical reaction force decreases. It is conceivable, then, that at some level of gravity the product of the vertical reaction force and the coefficient of friction will be insufficient to allow the man to develop the magnitude of the fore-and-aft shear force that is necessary to achieve his normal acceleration. He must therefore accelerate more slowly, by applying the available fore-and-aft shear force over a longer period of time. This is only one of the many possible mechanisms that may degrade walking performance at low-gravity levels.

In conclusion, it is restated that the magnitudes of the vertical force and the fore-and-aft shear force could be recorded as a function of time by the use of the force-measuring walkway. It is the author's contention that establishing the relationship between these forces at various gravity levels is prerequisite to determining the dominant mechanism in the degradation of low-gravity walking performance.

(The force-measuring walkway was not completed in time for use in this investigation.)

V. Low-Gravity, Time-Displacement Walking Experiment

Purpose

The purpose of this experiment was to quantitatively evaluate the motion parameters of the human gait under partial gravity conditions. It was desired to determine the acceleration and velocity of the subject as an overall indication of man's capability under reduced-gravity levels. Other gait parameters, such as the number of steps per minute, the step length, the swinging time, and the swing-to-support ratio were desired to permit an objective comparison of the low-gravity gait with the normal gait. It was hoped that evaluation of these later parameters would help explain the changes in the velocities and accelerations, and possibly define some of the capability limits.

Scope

This experiment was an end-view motion picture analysis of man walking at various reduced-gravity conditions. In the absence of the force-measuring platform, no attempt was made to evaluate the surface reaction forces.

Procedures and Technique

Aircraft Laboratory. Experiments were conducted in the C-131 B "weightlessness" airplane. The aircraft may be flown in a parabolic trajectory to induce any constant vertical acceleration from zero to 2.5 g, or may also be flown in a "decay maneuver" to produce transient accelerations throughout the same range. Each parabola provides approximately 15 seconds at the desired gravity condition.

The cargo compartment of the aircraft provides a working volume 73 in. high (at the centerline), 96 in. wide (at the floor level), and 25 ft long. The walls of the cabin are padded to reduce the impact shock of free-floating subjects. The floor is covered with a 2-in.-thick layer of Ensolite material (a stiff foam rubber also used on the floor of boxing rings.) Floodlamps are recessed into the walls to provide uniform illumination for photography.

Instrumentation. A 20-ft distance scale was fabricated by taping 0.2-ft-wide strips of dark cloth tape over the white Ensolite floor. The lateral edges of the tape were cut to form a distinct pattern to facilitate reduction of the photographic data. This scale, or walkway, was canted across the cabin floor at a 20° angle from the aircraft centerline to permit better photographic coverage (see Fig. 11).

FIG. 11 FIXED CAMERA VIEW OF WALKWAY AND SUBJECT B

Two 16-mm (BIA-type Bell and Howell and Bolex) cameras operating at 32 frames per second were used to photograph all walking trials. The first camera was hand-held and was used to record a "close up" view of the positions of the subject's feet upon the walkway pattern. The other camera was fixed at waist level at the forward end of the aircraft to provide an overall view of body posture and motion.

Subjects. The two walking subjects were selected on the basis of availability. Both were experienced in the aircraft maneuver; the first subject has flown more than 3,500 trajectories, and the second subject has flown approximately 500 trajectories. (See Appendix B for physical data.)

Run Procedures. The subject stood with his toes at the zero mark of the distance scale during the 2.5-g entry into the maneuver. The test monitor signaled the attainment of the desired gravity condition by dropping his hand. Cameras were started. The subject walked forward the length of the walkway, pivot turned, and proceeded aft to the starting position. Subjects were instructed to walk comfortably at whatever speed seemed optimal for maintaining body control at the given gravity condition.

Test Sequence. Twelve runs comprised a test series. Two runs at 1.0 g were first made while the aircraft was flying straight and level. Then, two runs were made successively at the gravity levels of 0.7 g, 0.4 g, 0.25 g, 0.17 g (lunar gravity), and 0.10 g. Upon completion, the subject repeated the series in reverse order. Thus, the subject performed a sequence of runs in both descending and ascending order of gravity conditions. The first series was completed July 1. A complete set, which consisted of a descending and ascending series, was completed by each subject on July 2. Including reruns of aborted runs, 62 runs were made.

Data

All data was recorded on motion picture film. Film identification numbers and reduced data are contained in Appendix B.

Data Reduction

Run Selection. Film from the five series was projected to determine the best runs for data reduction. All film was cataloged, noting any adverse conditions present that might affect the results. The best series for each subject was selected on the basis of complete film coverage and lack of adverse aircraft accelerations. Both series were run in the ascending order of gravity conditions. Film from both cameras was again reviewed to select the better run at each gravity level for each subject. Adverse aircraft acceleration and turbulence affected some runs. Contacts by the subject against the ceiling or walls of the aircraft invalidated other runs. If no adverse factors were noticed for either run at a given gravity condition, final selection was dictated by which run had the more complete coverage of the start, turn, and stop.

Data Log. The selected runs were projected frame by frame in a 16-mm projector equipped with a frame counter and an optional manual driving mechanism. Positions of the feet on the walkway surface were logged opposite the number of the frame in which the event occurred. Heel-strike (HS) was defined as the first frame in which the heel touched the walkway surface. Toe-off (TO) was defined as the last frame in which the toe remained in contact with the surface. The feet-together position (FT) was selected as the toe position of the stance foot as the swinging foot comes forward to pass it. Because of the forward inclination of the walking man, it is believed that the feet-together position represents adequately the projection of the body center of gravity on the walking surface. The feet-together position is therefore used as the X position of the moving subject.

Calculation of Parameters. It was possible to calculate the magnitudes of the various motion parameters from the position-versus-frame number information that was logged for each run. Only basic calculation techniques are stated here. Sample calculations may be found in Appendix B.

Time. At a constant camera speed of 32 frames per second, the time difference between frames is 0.03125 sec. The time in seconds between successive events is obtained by multiplying the difference in frames by the factor 0.03125; i.e.,

$$t\ (\text{seconds}) = 0.03125 \quad Fr \qquad (4)$$

Velocity. The forward velocity of the body mass can be obtained either by dividing the distance traveled between two or more feet-together positions (step relationships) by the time interval, or else by dividing the distance traveled during one or more strides by either or both feet by the respective time intervals (stride relationships). Both methods have comparable accuracy. However, if only velocity and acceleration are to be determined, the simplicity of the step method required far less reduction time.

Acceleration. The forward acceleration of the body mass can theoretically be derived by differentiating the velocity. It was thought that the acceleration could be obtained from the experimental data by dividing the changes in velocity during the first few steps by the respective time intervals. However, it was found that, with the exception of the lower gravity levels, the body accelerates to walking velocity in the first step. Unfortunately, inadequate film coverage of the first walking step of most runs precluded this calculation. Therefore, forward acceleration could only be inferred from the displacement-time graphs.

Step Frequency. As in the case of the velocity calculations, the step frequency (N, steps/minute) can be derived from either the step relationships or the stride relationships. The step frequency was reduced from the step relationships because the method was more direct.

Step Length. The step length (S) is equal to the distance between consecutive feet-together positions. It may also be obtained by dividing the stride length (S_{st}) in half, i.e.,

$$S = \frac{S_{st}}{2} \tag{5}$$

Phase Times and Ratios. The relative time spent during the swing and support phases of a walking cycle can be obtained only from the stride relationships. The swinging time (t_{sw}) is the time between toe-off and heel-strike of the same foot. The support time (t_{sup}) is the time between heel-strike and toe-off. The time for one stride or walking cycle (t_{st}) is equal to the sum of the swinging time and the support time; i.e.,

$$t_{st} = t_{sw} + t_{sup} \tag{6}$$

The ratio of any two phasic activities is obtained by dividing the time spent in one phase by the time spent in the other phase. The swing-to-support ratio is a primary parameter for the comparison of walking gaits.

Results

The magnitudes of the motion parameters that were determined from the low-gravity walking experiment are illustrated in Figs. 12 to 17. (See Table B-II for the tabulated values.) The respective curves are plotted separately for each subject. It was considered that averaging the data from only two subjects would only mask the fundamental relationships.

The distance versus time relationships at each gravity level are shown in Fig. 12 and Fig. 13. The distance traveled while walking aft was added to the distance traveled while walking forward to present a continuous curve. (With respect to the body axis system, all walking was forward.) The first, last, and turning steps at the ends of the walkway are shown in dashed lines. An overall distance and time scale was intentionally omitted because it was necessary to space the curves to avoid the confusion of overlapping data points.

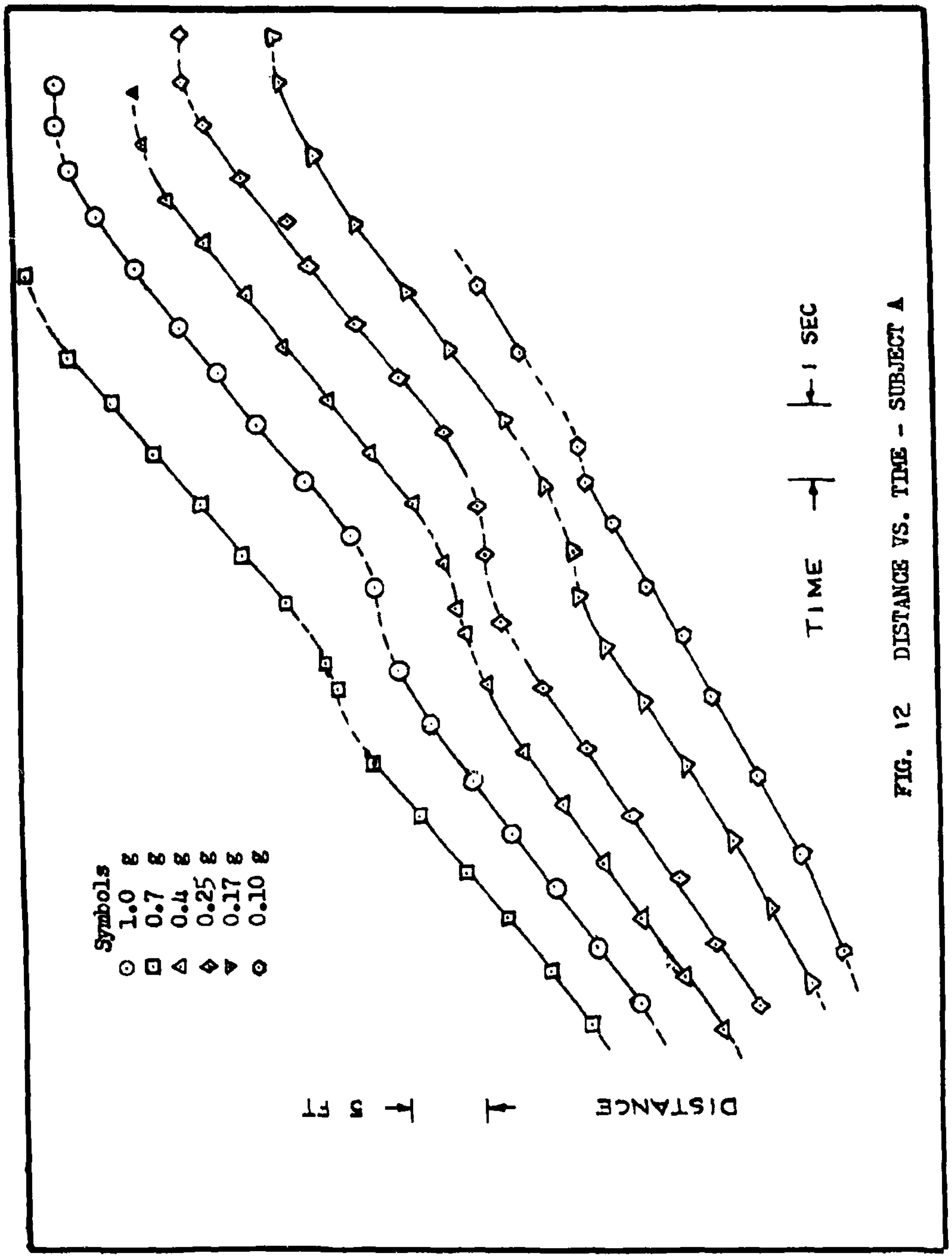


FIG. 12 DISTANCE VS. TIME - SUBJECT A

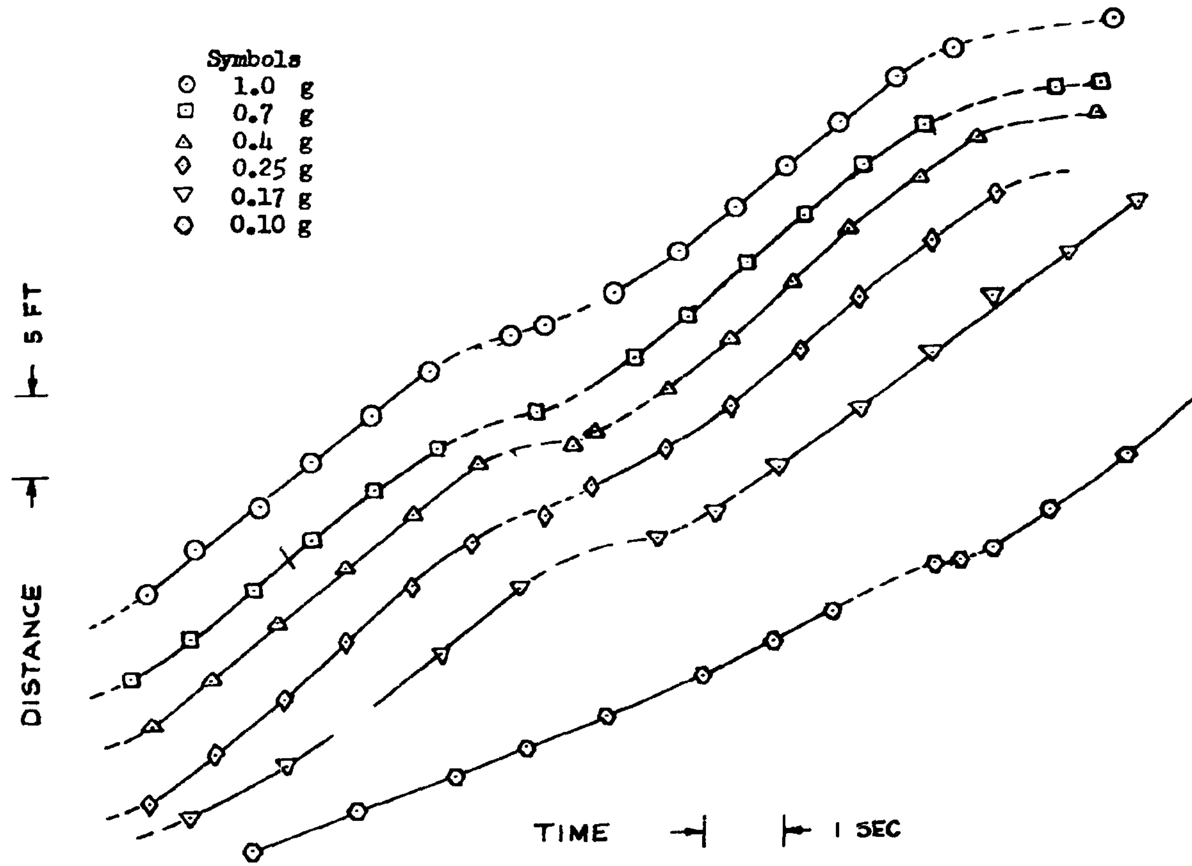


FIG. 13 DISTANCE VS. TIME - SUBJECT B

Only gross trends may be noticed from the distance versus time curves. The similarity of the curves tends to mask the changes in the various motion parameters. The values of the individual motion parameters may be shown more clearly by plotting them separately as a function of the gravity level. However, the similarity of the slopes of the curves indicates that the velocity varied little among the upper gravity levels. The linearity of the curves after the first step implies that the walking velocity is attained quickly and remains almost constant throughout the walk in the given direction. It is only at the lower gravity levels (as indicated by the slightly increasing slopes of the curves at the 0.17 and 0.10 gravity levels) that man continues to accelerate after his first walking step.

The step relationships of low-gravity walking are illustrated in Fig. 14. Both subjects walked with a velocity of 3.8 fps at the 1.0-g gravity level. Although the velocity was not a controlled variable, the subjects maintained a velocity within 10 per cent of the normal value throughout the descending range of gravity conditions to less than 0.25 g and 0.17 g, respectively. Velocity is proportional to the product of the step length and the step frequency. The factors contributing to the almost constant velocities varied between the subjects. It can be seen from the step length and the step frequency plots that

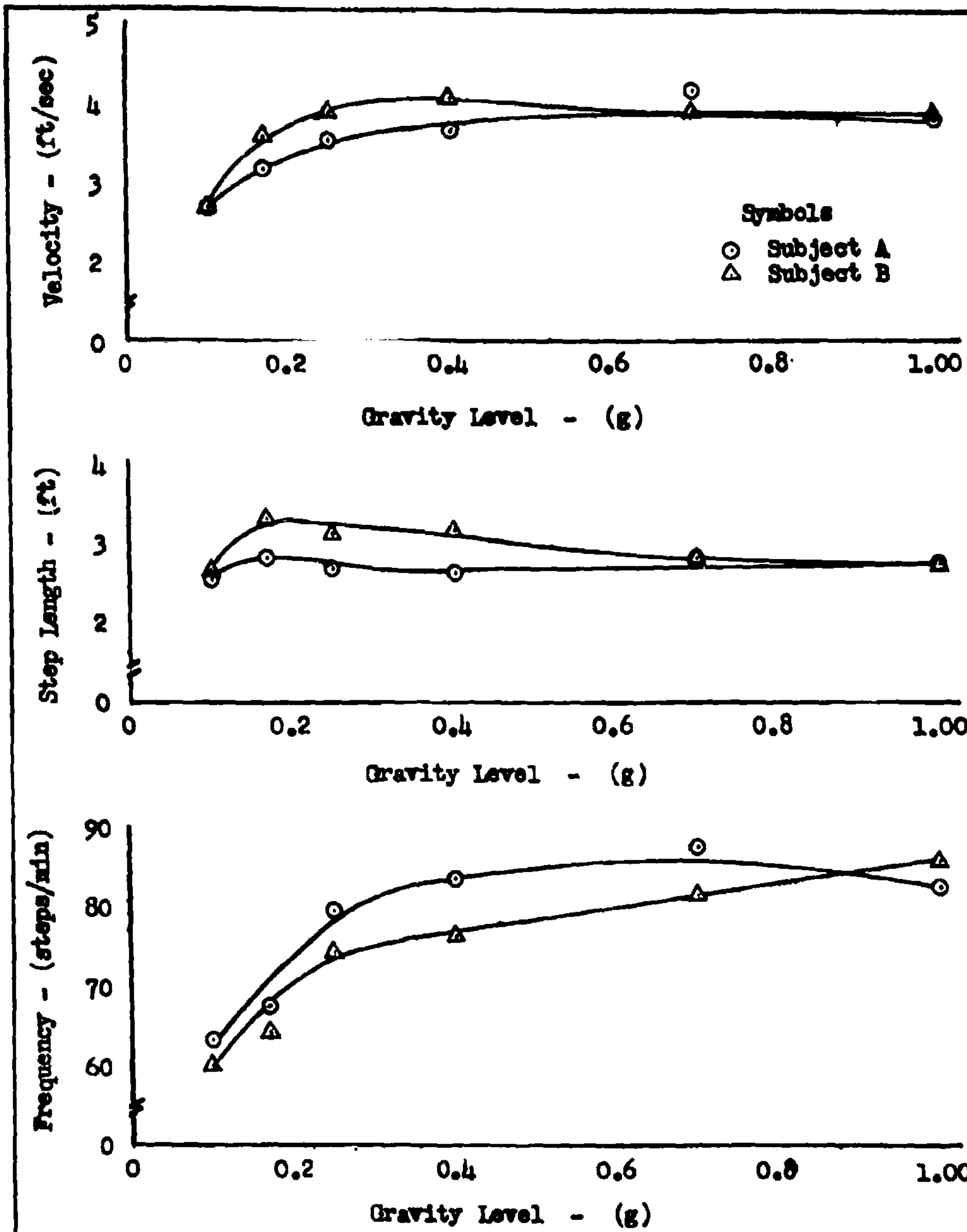


FIG. 14 STEP RELATIONSHIPS OF LOW-GRAVITY WALKING

Subject A minimized his velocity decrement by maintaining an almost constant step length and by increasing his step frequency. Subject B, however, increased his step length to offset the velocity decrease that would result from his decreased step frequency.

Although the curves are drawn smooth, the data points at the lower two gravity levels indicate that a discontinuity exists somewhere in the region of 0.2 g. This is most clearly shown on the frequency plot.

The times required for the respective phases of the walk are illustrated in Fig. 15. It may be seen that the support time for each subject remains essentially constant at gravity levels above 0.25 g although there is a fourteen per cent spread between the respective subjects. The variation in the stride time is chiefly due to the variation in the swing time. The influence of this spread is also evident in the stride time curves; however, the slope of the stride time curve is dominated by the influence of the swinging time.

The most consistent effect of the reduced-gravity conditions upon the motion parameters is evident in the swinging time plot. Despite fluctuations in the velocity, step length, and step frequency, the swinging time (with only one exception) increased at each successive reduction of the gravity level. This characteristic demonstrates the strong inverse dependence of the swinging time on gravity.

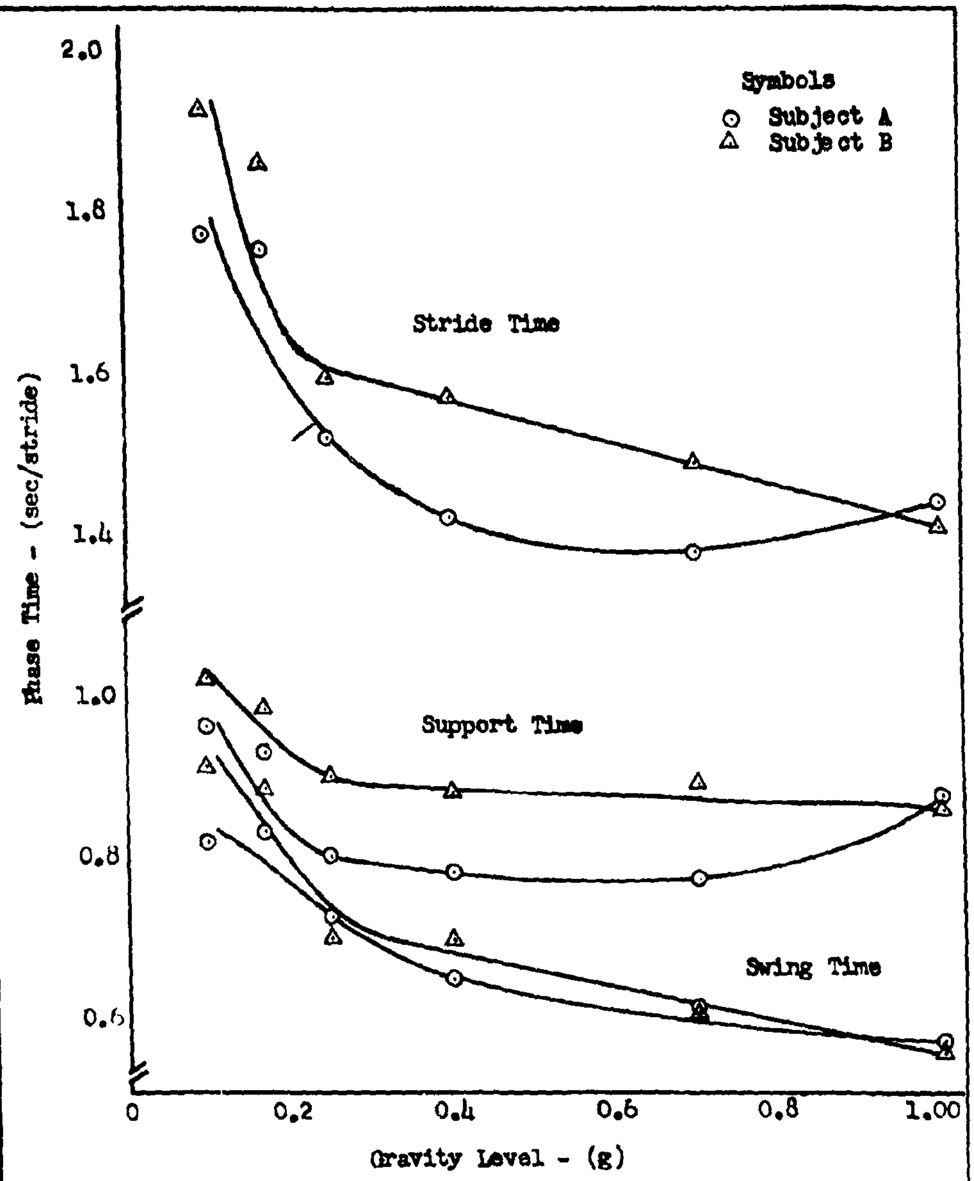


FIG. 15 PHASE TIMES OF LOW-GRAVITY WALKING

An attempt was made to find a mathematical expression that would approximate the swinging time curve. It was found that the swinging time varied inversely as the sixth root of the gravity level between the gravity levels of 0.25 g and 1.0 g, that is,

$$t_{sw} = t_{sw_{1.0\,g}} g^{-1/6} \qquad (7)$$

in the range

$$0.25 \leq g \leq 1.0$$

The discontinuity and spread of data points below the 0.25 g level degrades the validity of an approximate formula. However, the function

$$t_{sw} = t_{sw_{1.0\,g}} g^{-1/5} \qquad (8)$$

in the range

$$0.1 \leq g < 0.25$$

has the approximate slope and magnitude of the experimental data. A graph of these functions is superimposed on the swinging time curves in Fig. 16.

The phase ratios of the low gravity walk are shown in Fig. 17. The phase ratios provide a quantitative means of describing the manner of the gait. Under normal gravity conditions the swing-to-support ratio for normal walking is usually between 0.5 and 0.8 (Ref 5:7). The ratio increases with faster walking and approaches unity at the

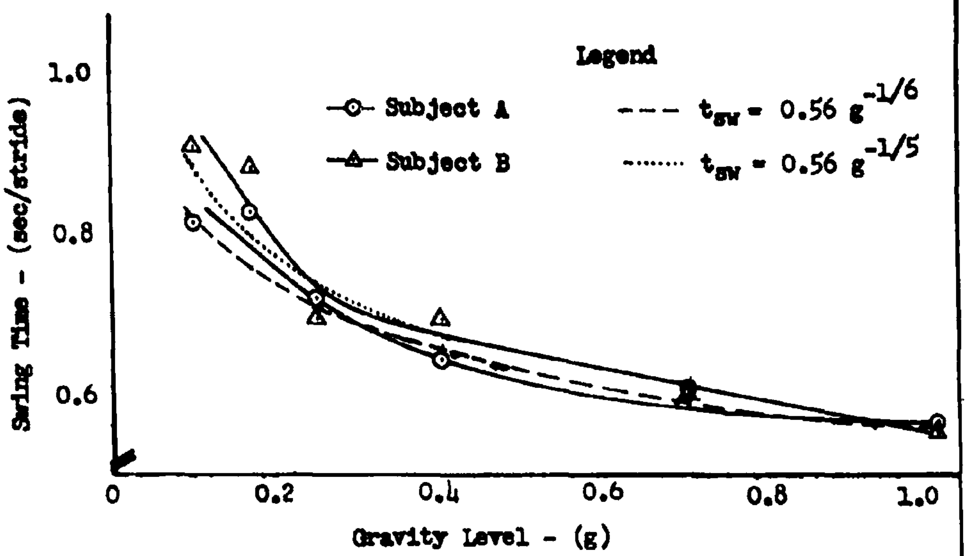


FIG. 16 MATHEMATICAL APPROXIMATION OF THE TIME OF SWING

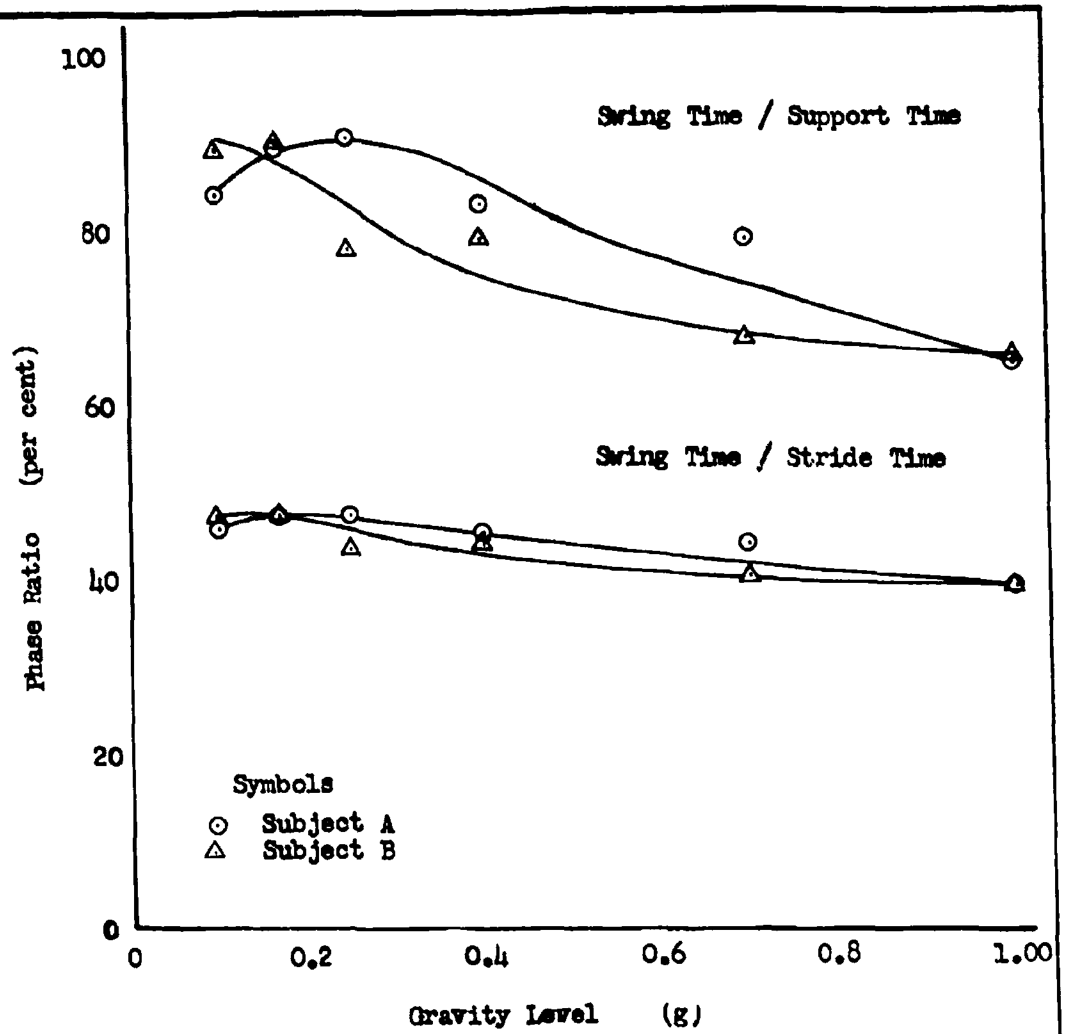


FIG. 17 PHASE RATIOS OF THE LOW-GRAVITY WALK

point of running. The swing-to-support ratio varied during this experiment from 0.65 at normal gravity to 0.90 at the lower gravity levels.

Most of the changes in walking motion are accomplished while both feet are on the surface. As the swing-to-support ratio increases, the duration of double support decreases and the subject has less control over his progression. The loss of body control indicated by the higher swing-to-support ratios at the lower gravity levels is clearly evident in the film record of the runs.

Based on the swing-to-support ratio, the low-gravity walking gait could be described as a fast walk. But, because velocity and step length do not increase, and the step frequency decreases, the low gravity walking gait is better described as a "fast walk in slow motion."

The swing-to-stride ratio was computed as an investigative parameter. It is concluded that this parameter presents no information that cannot be obtained more directly from the other parameters.

The results of the experimental program may be summarized as follows:

1. With the exception of the acceleration, the experimental method was adequate to quantitatively evaluate the motion variables of man's low gravity walking performance.

2. Both subjects maintained a velocity within 10 per cent of their 1.0-g walking velocity (3.8 fps) at gravity levels between 1.0 g and 0.25 g or below.

3. The increase in swinging time was the most consistent effect of the reduction of gravity. The swinging time may be approximated by the expression

$$t_{sw} = t_{sw_{1.0\,g}} g^{-1/6} \tag{7}$$

within the range

$$0.25 \leq g \leq 1.0$$

and

$$t_{sw} = t_{sw_{1.0\,g}} g^{-1/5} \tag{8}$$

in the range

$$0.1 \leq g < 0.25$$

4. The high value (0.90) of the swing-to-support ratio at the lower gravity levels indicates the reduction in control that the subject has over his motion.

5. The curves of almost all variables plotted as a function of gravity level exhibit a discontinuity in the vicinity of 0.2 g. This tends to substantiate earlier observations that the lower gravity limit for acceptable walking is in this region.

6. Qualitatively, the gait at low gravity levels appears to be a "fast walk in slow motion."

VI. Concluding Statements and Recommendations for Future Study

Concluding Statements

The experimental methods for the study of low-gravity walking may be divided into two categories according to the basis of the experimental approach:

1. Experimental motion measurement and analysis, and
2. Experimental force measurement and analysis.

Walking performance characteristics such as velocity, step length, step frequency, and the phase times and ratios, may be quantitatively determined by a motion analysis. The acceleration of the subject could probably be determined with only minor improvements in the test procedure. However, motion analysis yields little information that concerns the reasons for the discontinuity in the walking performance in the region of 0.2 g.

A force analysis which includes the measurement of the surface reaction forces is the most direct approach for interpreting the causes of the degraded walking performance at the lower gravity levels. The proposed force-measuring walkway was conceived as the best means of recording these forces.

A summary of the results of the experimental motion analysis experiment is contained in Chapter V. The quantitative magnitudes of the various parameters as a function of gravity level are listed in Table B-II and are plotted in Figs. 12 to 17. It must be emphasized that this experiment was conducted in a "shirtsleeve" environment by evaluating only a limited number of runs of two unencumbered subjects during relatively short periods of partial-gravity conditions. It is believed that the results accurately describe the exhibited walking performance. However, extrapolation of the results beyond the range of the measured gravity levels, or to the performance of a man encumbered by the addition of mass or the restrictions of a pressure suit, is deemed neither valid nor advisable.

Recommendations for Future Study

It is recommended that this investigation be continued to accomplish the following objectives:

1. Quantitatively evaluate the walking performance of a pressure-suited subject under reduced-gravity conditions.

2. Quantitatively evaluate the effect of the addition of mass to the walking subject as a function of the gravity level.

3. Evaluate the acceleration capabilities of the walking subject as a function of gravity level.

4. Establish the relationships between the vertical reaction force and the fore-and-aft shear force as a function of gravity level.

5. Identify and evaluate the cause of the degraded walking performance in the region of 0.2 g.

6. Refine the estimates of the lower limit of gravity at which man can walk unaided.

7. Evaluate the combination of any of the previous objectives.

Pressure-suited behavior, the effect of the addition of mass, acceleration capabilities, and estimates of the lower gravity limit may be determined by the given motion analysis technique. The relationships between the surface reaction forces, and the cause of the degraded low-gravity walking performance may only be determined by use of the force-measuring walkway or another investigative technique. A combination of both force and motion techniques would facilitate the achievement of objectives 2, 3, and 6.

Bibliography

1. Bresler, B., and F. R. Berry. Energy and Power in the Leg During Normal Level Walking. Report (Series 11, Issue 15) to the Advisory Committee on Artificial Limbs, National Research Council. Berkeley, California: Institute of Engineering Research, University of California, May 1951.

2. Bresler, B., and J. P. Frankel. "The Forces and Moments in the Leg During Level Walking." Transactions of the American Society of Mechanical Engineers, 72:27-36 (January 1950).

3. Eberhart, H. D., and V. T. Inman. "An Evaluation of Experimental Procedures Used in a Fundamental Study of Human Locomotion." Annals New York Academy of Sciences, 51:1213-1228 (January 1951).

4. Eshbach, Ovid W. Handbook of Engineering Fundamentals (Second Edition). New York: John Wiley & Sons, Inc., 1952.

5. Finley, F. R., and P. V. Karpovich. Electrogoniometric Analysis of Normal and Pathological Gaits. Report. Springfield, Massachusetts: Department of Physiology, Springfield College, May 1961.

6. Fundamental Studies of Human Locomotion and Other Information Relating to Design of Artificial Limbs. Subcontractor's Report (Serial No. CAL 5) to the Committee on Artificial Limbs, National Research Council. Berkeley, California: Institute of Engineering Research, University of California, June 1947. (2 Volumes).

7. Gollnick, P. D., P. V. Karpovich, and J. Sharpe. Electrogoniometric Study of Locomotion and of Some Athletic Movements. Report. Springfield, Massachusetts: Department of Physiology, Springfield College, September 1961.

8. Investigations with Respect to the Design, Construction and Evaluation of Prosthetic Devices. Report No. 80.10 to the Special Devices Center, Office of Naval Research. New York: College of Engineering, New York University, 1949 (2 Volumes).

9. Levens, A. S., V. T. Inman, and J. A. Blosser. "Transverse Rotation of the Segments of the Lower Extremity in Locomotion." The Journal of Bone and Joint Surgery, 30-A:859-872 (October 1948).

10. Loret, Benjamin J. Optimization of Manned Orbital Satellite Vehicle Design with Respect to Artificial Gravity. Thesis, published as ASD Technical Report 61-688. Wright-Patterson AFB, Ohio: Aeronautical Systems Division, December 1961.

11. NASA Life Sciences Data Book. Handbook. Yellow Springs, Ohio: Webb Associates, June 1962.

12. Perry, C. C., and H. R. Lissner. The Strain Gage Primer (Second Edition). New York: McGraw-Hill Book Company, Inc., 1962.

13. Proposal for Biokinetics Study of Humans in Simulated Weightless Condition. Report No. ER-2885. Cockeysville, Maryland: Aircraft Armaments, Inc., November 1962.

14. Ryker, N. J., Jr. Glass Walkway Studies of Normal Subjects During Normal Level Walking. Report (Series 11, Issue 20) to the Advisory Committee on Artificial Limbs, National Research Council. Berkeley, California: Institute of Engineering Research, University of California, January 1952.

15. Saunders, J. B. deC. M., V. T. Inman, and H. D. Eberhart. "The Major Determinants in Normal and Pathological Gait." The Journal of Bone and Joint Surgery, 35-A:543-558 (July 1953).

16. Schwartz, R. P., and A. L. Heath. "The Definition of Human Locomotion on the Basis of Measurement." The Journal of Bone and Joint Surgery, 29:203-214 (January 1947).

17. Sharp, Earl D., and Charles W. Sears. Walking Under Zero-Gravity Conditions Using Velcro Material. AMRL Memorandum P-23. Wright-Patterson AFB, Ohio: 6750th Aerospace Medical Research Laboratories, January 1963.

18. Simons, John C. Walking Under Zero-Gravity Conditions. WADC Technical Note 59-327. Wright-Patterson AFB, Ohio: Wright Air Development Center, October 1959.

19. Steindler, Arthur. Mechanics of Normal and Pathological Locomotion in Man. Springfield, Illinois: Charles C. Thomas, 1935.

Appendix A

Calculation of the Atmospheric Drag on a Walking Man

The magnitude of the atmospheric drag on the body may be computed by the formula

$$D = \frac{1}{2} \rho V^2 \; S \; C_D \qquad \text{(A-1)}$$

where

D = drag force, lb_f

ρ = atmospheric density, $slug/ft^3$

V = velocity of the body through the medium, ft/sec

S = projected frontal area of the body, ft^2

C_D = drag coefficient

The average walking velocity is 5 ft/sec. However, the forward swinging arm and leg will progress at approximately twice this velocity. Therefore, since drag varies as the square of velocity, this effect is accounted for by assuming that the swinging velocity is 10 ft/sec and the swinging area is one-third the projected frontal area of the entire body. The projected frontal area of the Air Force mean man is 6 ft^2 (Ref 11:152). The coefficient of drag is assumed to be that of a flat plate positioned perpendicularly to the airstream, i. e., $C_D = 2.0$. Assuming sea-level density, the total atmospheric drag on the body is

$$D = \frac{1}{2} (0.002378)(6.0)(2.0) \left[\frac{1}{3} (10)^2 + \frac{2}{3} (5)^2 \right]$$

$$= 0.71 \; lb_f$$

The body assumes a forward inclination to resist the rearward moment caused by the drag force. For the purpose of this calculation, the drag force is assumed to act at the body center of mass. The magnitude of the forward inclination for a 160-pound man is computed to be

$$\theta = \tan^{-1}\left(\frac{0.71}{160}\right) = \tan^{-1}(0.00446) = 0.25^{\circ} \qquad \text{(A-2)}$$

where θ is the change in the angle of inclination measured from the vertical.

The vertical position of the center of mass is assumed to be 40 inches above the point of support. The corresponding shift of the center of mass forward from the "no-wind" point of support is

$$\begin{aligned} X &= 40 \sin\theta \\ &= 40 \sin(0.25^{\circ}) \\ &= 0.178 \text{ in.} \end{aligned} \qquad \text{(A-3)}$$

It should be noted that the assumed values are grossly conservative. Therefore, the author considers that the effect of drag on the body at normal walking speeds is negligible.

Appendix B

Experimental Data

Table B-I

Subjects' Physical Data

Subject	Age	Height (in.)	Weight (lb)
A	24	67-1/2	170
B	21	72	173

Data Log, Subject A

The following six runs were performed by Subject A on July 2, 1963, and were selected as the better of the two runs at each gravity condition. The series (number 3) was conducted in ascending order of gravity levels. The runs appear in the following order on the respective films:

Frame No.	Run No.	Gravity Level	Film No.
6,600	27	0.10	Hand Camera (Roll 2,3,4) w/o 9023(623)
7,400	29	0.17	" "
8,793	32	0.25	" "
9,831	34	0.40	" "
10,800	36	0.70	" "
0	37	1.00	Hand Camera, 2 Jul 63, w/o 9023(623)

Run 27, 0.10 g, Subject A, 2 Jul 63, Start Frame 6,600.									
Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	---	0.0				
			LHS	---	~0.7				
RTO	~18	0.0							
						FT	31	1.6	
RHS	52	3.8	LTO	60	1.6				
						FT	72	4.4	
			LHS	89	6.5				
RTO	93	4.2							
						FT	105	7.2	
			LTO	125	7.2				
RFF	128	!19.3							Almost floating
						FT	140	10.0	
RTO	155	10.0	LHS	155	10.9				
						FT	166	11.9	
RHS	177	13.3	LTO	177	12.0				
						FT	186	14.1	
			LHS	198	15.5				Decelerative HS
RTO	204	14.2							
						FT	214	16.3	
			LTO	223	16.4				
RHS	226	17.5							
						FT	231	18.2	
						CG	250	18.5	Last frame before missing frames
			LHS	260	15.5				
RTO	259	17.4							
						FT	270	14.6T	
RHS	290	12.8	LTO	290	14.6				
						FT	300	12.0	
RTO	310	12.0	LHS	317	9.5				
RTS	335	9.7							Fall forward, hand contacted floor

Run 29 0.17 g Subject A 2 Jul 63 Start Frame 7,400

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	---	0.0				
			LHS	---	2.0 T 1.1 M				
RTO	0	0.0							
						FT	8	2.0	
RHS	25	4.0							
			LTO	27	2.1				
						FT	39	4.5	
			LHS	53	6.2				
RTO	57	4.6				FT	69	7.1	
RHS	87	9.3							
			LTO	89	7.3				
						FT	100	10.0	
			LHS	114	11.9				
RTO	116	10.1							
						FT	128	12.8	
RHS	140	14.4							
			LTO	140	12.9				
						FT	151	15.3	
			LHS	163	16.1				
RTO	164	15.3							
						FT	173	17.0	
RFF	~185	18.3				CG	185	17.5	
									Frames missing
						FT	187	15.8	
RTO	205	16.8	LHS	205	15.1				
						FT	215	14.2	
RHS	226	12.0							
			LTO	227	14.1				
						FT	241	10.8	
			LHS	255	8.8				
RTO	256	10.9							
						FT	271	7.9	
RHS	283	5.6							
			LTO	287	7.9				
						FT	300	4.6	
			LHS	315	2.8				
RTO	318	4.6							
						FT	329	1.9	
RHS	343	0.4							
			LTO	344	2.0	FT	359	-0.4	
			LFF	378	-0.2				
						CG	378	-0.6	

Run 32 0.25 g Subject A 2 Jul 63 Start Frame 8,793

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
						FT	0	-1.2	
RHS	10	0	L						
			LTO	19	-1.2				
						FT	30	0.8	
			LHS	43	2.6				
RTO	48	0.8							
						FT	57	3.6	
RHS	70	5.6							
			LTO	71	3.6				
						FT	84	6.0	
			LHS	98	8.4				
RTO	99	6.0							
						FT	111	9.1	
RHS	126	11.1							
			LTO	128	9.1				
						FT	139	12.0	
			LHS	152	13.8				
RTO	155	12.0							
						FT	165	14.6	
RHS	178	16.6							
			LTO	181	14.6				
						FT	193	17.5	
			LHS	208	19.2				
RTO	210	17.6							Right foot pivot
						CG	223	18.4	
RFF	---	17.8 18.7							
			LTO	236	19.1	FT	243	17.8	
			LHS	263	16.5				
RTO	266	17.8							
						FT	275	15.7	
RHS	286	13.8							
			LTO	288	15.7				
						FT	298	12.8	
			LHS	309	10.9				
RTO	311	12.9							
						FT	322	10.1	
RHS	335	7.7							
			LTO	335	10.1				
						FT	346	7.0	
			LHS	355	5.6				
RTO	359	6.9							
						FT	366	4.8	
RHS	378	3.4							

Run 32, continued

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	379	4.8				
						FT	384	2.9	
			LHS	395	1.5				
RTO	398	2.8							
						FT	406	0.5	
RHS	413	-0.3							
			LTO	416	0.5				
						FT	424	-1.0	
RFF	432	-1.4							
						CG	445	-1.0	
						CG	460	-0.5	

Run 34 0.40 g Subject A 2 Jul 63 Start Frame 9,831

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	---	----				
			LHS	8	1.3				
RTO	11	0.0							
						FT	21	2.1	
RHS	31	4.0							
			LTO	35	2.3				
						FT	44	4.6	
			LHS	55	6.6				
RTO	58	4.7							
						FT	68	7.3	
RHS	79	9.2							
			LTO	82	7.4				
						FT	92	9.8	
			LHS	104	11.7				
RTO	107	9.8							
						FT	117	12.4	
RHS	128	14.0							
			LTO	131	12.5				
						FT	140	14.9	
			LHS	152	16.4				
RTO	156	15.0							
						FT	167	17.2	
RHS	175	18.2							

Run 34, continued

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	180	17.3				
						FT	191	19.1	
			LFF←	195	19.4				
						CG	200	19.1	
RTO - RHS ←---									
			LTO	215	19.2				
						FT	219	18.3	
			LHS	232	17.2				
RTO	237	18.3							
						FT	245	16.2	
RHS	254	14.6							
			LTO	257	16.2				
						FT	266	13.6	
			LHS	277	12.0				
RTO	279	13.7							
						FT	289	10.9	
RHS	300	9.2							
			LTO	301	11.0				
						FT	312	8.1	
			LHS	321	6.6				
RTO	325	8.2							
						FT	335	5.7	
RHS	344	4.0							
			LTO	346	5.6				
						FT	356	2.9	
			LHS	366	1.2				
RTO	370	2.9							
						FT	380	0.5	
RHS	390	-0.4							
			LTO	390	0.4				
						CG	399	-1.0	
						FT	399	-1.2	
			LHS	407	-1.8				
RTO	411	---							
						CG	420	-1.5	

Run 36 0.70 g Subject A 2 Jul 63 Start Frame 10,800

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	0	0.0				
			LHS	8	1.9				
RTO	15	0.0							
						FT	23	2.7	
RHS	33	4.5							
			LTO	36	2.7				
						FT	45	5.3	
			LHS	57	7.5				
RTO	59	5.3							
						FT	68	8.1	
RHS	79	102							
			LTO	81	8.2				
						FT	89	11.0	
			LHS	99	13.1				
RTO	103	11.1							
						FT	112	13.9	
RHS	123	16.1							
			LTO	125	14.0				
						FT	135	16.9	
			LHS	149	19.0				
RTO	151	16.9							
RFF	166	19.0							
						CG	167	19.0	
			LTO	172	19.4				
						FT	177	18.1	
			LHS	191	16.5				
RTO	194	18.1							
						FT	203	15.6	
RHS	214	13.8							
			LTO	217	15.6				
						FT	224	12.9	
			LHS	236	11.0				
RTO	238	12.9							
						FT	246	10.1	
RHS	257	8.2							
			LTO	258	10.1				
						FT	267	7.2	
			LHS	277	5.2				
RTO	279	7.3							
						FT	289	4.4	
RHS	298	2.4							
			LTO	302	4.4				
						FT	310	1.5	
			LHS	320	-0.3				

Run 36, continued									
Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
RTO	324	1.6							
RFF	340	-0.9							
						CG	344	-1.2	

Run 37 1.00 g Subject A 2 Jul 63 Start Frame 0									
Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	---	----				
			LFF	---	1.6				
RTO	23	0.0							
						FT	32	2.4	
RHS	43	4.5							
			LTO	47	2.4				
						FT	56	5.2	
			LHS	67	7.2				
RTO	72	5.2							
						FT	81	8.0	
RHS	90	9.9							
			LTO	95	8.0	FT			
						FT	104	10.6	
			LHS	114	12.6				
RTO	119	10.6							
						FT	127	13.3	
RHS	136	17.1							
			LTO	14.1	13.4				
						FT	151	16.0	
			LHS	159	17.1				
RTO	166	16.0							
						FT	174	18.0	
RHS	181	18.9	LTO	188	18.0				
			LFF	205	20.0				
						CG	210	19.5	
									Frames missing RTO-RHS-LTO-LHS
RHS	---	18.0							
			LTO	---	19.5				
						FT	216	17.0	
			LHS	226	15.5				

Run 37, continued

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
RTO	231	17.0							
						FT	240	14.5	
RHS	251	12.9							
			LTO	255	14.5				
						FT	263	12.0	
			LHS	272	10.2				
RTO	278	12.0							
						FT	286	9.4	
RHS	294	7.6							
			LTO	299	9.3				
						FT	307	6.7	
			LHS	317	5.1				
RTO	322	6.7							
						FT	330	4.1	
RHS	339	2.5							
			LTO	344	4.3				
						FT	352	1.6	
			LHS	360	0.8				
RTO	367	1.6							
						FT	378	-0.2	
RHS	379	-0.4							
			LTO	384	-0.3				
						CG	392	-1.0	Last frame

Data Log, Subject B

The following six runs were performed by Subject B on July 2, 1963, and were selected as the better of the two runs at each gravity condition. The series (number 5) was conducted in ascending order of gravity levels. The runs appear in the following order on the film:

Frame No.	Run No.	Gravity Level	Film No.
8,300	56	0.25 g	Hand Camera, 2 Jul 63, w/o 9023(623)
8,800	57	0.40 g	" "
10,650	60	0.70 g	" "
11,394	62	1.00 g	" "
13,792	52	0.10 g	" "
14,521	53	0.17 g	" "

Run 52 0.10 g Subject B 2 Jul 63 Start Frame 13,792									
Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	---	----				
			LHS	18	1.1				
RTO	31	0.0							
						FT	44	2.1	
RHS	55	2.9							
			LTO	72	2.2				
						FT	85	4.5	
			LHS	110	5.8				
RTO	113	4.3							
						FT	124	6.6	
RHS	139	7.5	LTO	146	6.6				
						FT	152	8.3	
			LHS	170	9.3				
RTO	174	8.2							
						FT	185	10.2	
			LTO	206	10.3				
RHS	208	11.6							
						FT	222	12.6	
			LHS	234	13.7				
RTO	239	12.6							
						FT	250	14.6	
RHS	260	15.4							
			LTO	264	14.7				
						FT	273	16.4	
			LHS	288	19.5				
RTO	295	----							
RTS↑	314	18.3				CG	314	19.3	
						CG	324	19.2	
			LTO↑	334	19.3				
						FT	337	18.4	
			LHS↑	349	17.2				
RTO	352	18.3							
						FT	360	16.1	
RHS	373	13.8							
			LTO	377	16.0	FT	390	12.8	
			LHS	407	10.1				
RTO	406	12.8				FT	420	9.0	
RHS	433	6.6							
			LTO	435	8.9				
						FT	448	5.6	
RTO	466	5.6							

Run 52, continued

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LHS	468	2.9				
						FT	483	1.8	
RHS	496	0.5							
			LTO	499	1.8				
						FT	516	-0.5	
						CG	583	-0.6	Pivot stop

Run 53 0.17 g Subject B 2 Jul 63 Start Frame 14,521

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
RHO	2	-1.0							
RTO	5	0.0							
RHS	33	1.7							
			LTO	42	0.0				
						FT	50	2.4	
			LHS	69	4.0				
RTO	75	2.4				FT	88	5.4	
RHS	105	7.6							
			LTO	106	5.5				
						FT	118	8.3	
RTO	135	8.7							
			LHS	136	11.4				
						FT	150	12.2	
RHS	166	15.2	LTO	166	12.2				
						FT	180	16.2	
			LHS	200	19.2				
RTO	202	16.6							Slipping
RHS	---	----				CG	235	19.2	
RTO†	235	18.0							
RTS	248	17.8							
			LTO	252	19.2				
						FT	258	17.8	
			LHS	269	16.0				
RTO	273	17.8				FT	284	14.9	
RHS	301	12.6							

Run 53, continued									
Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	303	15.0				
						FT	316	11.0	
			LHS	331	9.2				
RTO	331	11.5							
						FT	344	8.1	
RHS	354	5.7							
			LTO	356	8.1	FT	367	4.6	
			LHS	382	2.0				
RTO	385	4.5							
						FT	397	2.0	
RHS	411	-0.4							
			LTO	---	0.8				
						FT	426	-1.0	
			LHS	454	0.0				Hand contact during turn-around

Run 56 0.25 g Subject B 2 Jul 63 Start Frame 8,300									
Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LHS	19	1.7				
RTO	24	0.0							
						FT	35	2.6	
RHS	48	4.6							
			LTO	53	2.6				
						FT	61	5.6	
			LHS	77	8.1				
RTO	81	5.6							
						FT	89	8.9	
RHS	101	11.4							
			LTO	103	9.1				
						FT	112	12.4	
			LHS	125	14.8				
RTO	127	12.4							
						FT	139	15.8	
RHS	151	17.6							
			LTO	153	15.8				
						FT	163	18.5	
R	---	18.1							Right foot pivot
						CG	192	20.0	

Run 56, continued

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	202	20.0				
						FT	210	18.1	
			LHS	226	16.8				
RTO	230	18.1							
						FT	239	15.8	
RHS	251	14.2							
			LTO	257	15.9				
						FT	265	13.2	
			LHS	279	11.0				
RTO	282	13.2							
						FT	292	9.9	
RHS	303	7.6							
			LTO	306	10.0				
						FT	316	6.6	
			LHS	330	4.1				
RTO	333	6.6							
						FT	344	3.2	
RHS	355	1.2							
			LTO	358	3.3				
						FT	370	0.3	
						CG	413	-0.6	Hands on back wall both feet pivoting

Run 57 0.40 g Subject B 2 Jul 63 Start Frame 8,800

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LHO	2	0.0				
			LTO	6	0.0				
			LHS	20	1.7				
RTO	26	0.0							
						FT	36	2.5	
RHS	46	4.5							
			LTO	51	2.3				
						FT	60	5.3	
			LHS	72	7.8				
RTO	75	5.4							
						FT	86	8.8	
RHS	100	11.3							

Run 57, continued

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	102	8.9				
						FT	112	12.1	
			LHS	126	14.4				
RTO	128	12.2							
						FT	139	15.4	
RHS	151	17.6							
			LTO	155	15.4				
						FT	165	18.5	
			LHS	173	19.8				
RTO	178	18.4				CG	202	19.5	Right foot pivoting
			LTO	204	19.6				
						FT	211	18.5	
			LHS	224	17.8				
RTO	234	18.5							
						FT	240	16.2	
RHS	---	14.0							Frame blurred
			LTO	255	16.2				
						FT	265	13.0	
			LHS	277	10.6				
RTO	280	13.0							
						FT	290	9.6	
RHS	301	7.6							
			LTO	303	9.7				
						FT	313	6.4	
			LHS	326	4.4				
RTO	330	6.4							
						FT	340	3.4	
RHS	348	1.9							
			LTO	353	3.4				
						FT	363	0.8	
			LHS	375	-1.0				
RTO	376	0.8							
RTS	398	0.0							
						CG	410	-0.6	

Run 60 0.70 g Subject B 2 Jul 63 Start Frame 10,650									
Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	0	0.0				First steps fuzzy,
			LHS	12	1.8				Camera not focused
RTO	18	0.0							
						FT	27	2.6	
RHS	39	4.6							
			LTO	42	2.6				
						FT	51	5.0	
			LHS	65	7.4				
RTO	68	5.1							
						FT	76	8.1	
RHS	87	10.3							
			LTO	91	8.2				
						FT	99	11.2	
			LHS	109	13.2				
RTO	115	11.2							
						FT	124	14.1	
RHS	134	15.8							
			LTO	140	14.1				
						FT	148	16.7	
			LHS	160	18.4				
RTO	165	----							
RFF	---	18.3				CG	188	18.9	
			LTO	191	18.9				
			LHS	213	17.7				
RTO	218	18.3							
						FT	226	15.6	
RHS	233	14.0							
			LTO	239	15.6				
						FT	247	13.0	
			LHS	258	10.8				
RTO	262	13.0							
						FT	271	9.8	
RHS	281	8.0							
			LTO	284	9.8				
						FT	293	7.0	
			LHS	304	5.0				
RTO	308	6.9							
						FT	317	3.8	
RHS	328	2.4							
			LTO	332	3.9				
						FT	341	1.4	
			LHS	351	0.0				
RTO	357	---				CG	393	-0.8	Right Foot Pivot

Run 62 1.00 g Subject B 2 Jul 63 Start Frame 11,394

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
			LTO	5	0.0				
			LHS	18	1.9				
RTO	24	0.0							
						FT	33	2.8	
RHS	41	4.8							
			LTO	46	2.8				
						FT	53	5.7	
			LHS	65	7.6				
RTO	69	5.6							
						FT	78	8.3	
RHS	86	10.2							
			LTO	91	9.4				
						FT	99	11.0	
			LHS	108	12.9				
RTO	~113	11.1							
						FT	122	13.7	
RHS	131	15.6							
			LTO	137	13.8				
						FT	145	16.5	
			LHS	153	18.6				
RTO	160	16.6							
RFF	172	17.8				CG	177	18.6	
			LTO	186	19.2				
						FT	191	17.8	
			LHS	205	17.1				
RTO	212	17.9							
						FT	219	16.0	
RHS	232	14.5							
			LTO	235	16.0	FT	244	13.5	
			LHS	254	11.9				
RTO	258	13.5							
						FT	266	10.8	
RHS	274	9.3							
			LTO	279	10.8				
						FT	287	8.3	
			LHS	295	6.5				
RTO	300	8.3							
						FT	309	5.5	
RHS	318	3.9							
			LTO	323	5.4				
						FT	331	2.9	
			LHS	340	2.0				

Run 62, continued

Right Foot			Left Foot			Feet Together			Remarks
Phase	Fr	Pos	Phase	Fr	Pos	Phase	Fr	Pos	
RTO	347	2.8							
						FT	354	1.0	
RHS ↗	364	---							
			LTO	367	0.8				
			LFF →	387	-1.3				
RTO →	390	-0.6							
RFF ↑	412	0.0							
						CG	417	-0.7	

Sample Calculations

The following sample calculations are illustrated for Run 37, Subject A, at the 1.0-g gravity level.

The velocity, length of step, and step frequency are determined from the step (feet-together) relationships. The first and last steps in each direction are disregarded because of acceleration and deceleration. The computed values are thus the average values for the "steady state" portion of the walk.

Walking forward, the subject in five steps traveled from 2.4 ft at frame number 32 to 16.0 ft at frame number 151. The distance traveled is given by the equation

$$d = d_2 - d_1 \tag{B-1}$$

$$d_{fwd} = 16.0 - 2.4$$

$$= 13.6 \text{ ft}$$

At a camera speed of 32 frames per second the frame numbers may be converted into time by the equation

$$t = 0.03125 \ Fr \tag{B-2}$$

The time for the five steps is therefore

$$t = 0.03125 \ (Fr_2 - Fr_1) \tag{B-3}$$

$$t_{fwd} = 0.03125 \ (151 - 32)$$

$$= 3.72 \text{ sec}$$

The velocity is determined from the equation

$$V = \frac{d}{t} \tag{B-4}$$

$$V_{fwd} = \frac{13.6}{3.72}$$
$$= 3.66 \text{ ft/sec}$$

The step length is obtained by dividing the distance traveled by the number of steps

$$S = \frac{d}{\text{no. of steps}} \qquad \text{(B-5)}$$

$$S_{fwd} = \frac{13.6}{5}$$
$$= 2.72 \text{ ft}$$

The step frequency is equal to the number of steps divided by the difference in time. For comparison with published values the frequency is converted from steps per second to steps per minute.

$$N = \frac{\text{number of steps}}{t} \times 60 \qquad \text{(B-6)}$$

$$N_{fwd} = \frac{5\ (60)}{3.72}$$
$$= 80.6 \text{ steps/min}$$

Walking aft, the subject in five steps traveled from 17.0 ft at frame number 216 to 4.1 ft at frame number 330. The values of the parameters for the aft walk are therefore

$$t_{aft} = 0.03125\ (330 - 216)$$
$$= 3.56 \text{ sec}$$

$$d_{aft} = 17.0 - 4.1$$
$$= 13.9 \text{ ft}$$

$$V_{aft} = \frac{13.9}{3.56}$$

$$= 3.90 \text{ ft/sec}$$

$$S_{aft} = \frac{13.9}{5}$$

$$= 2.78 \text{ ft}$$

$$N_{aft} = \frac{5\ (60)}{3.56}$$

$$= 84.2 \text{ steps/min}$$

The average value of a parameter at the gravity level is equal to one-half the sum of the respective forward and aft values, i.e.,

$$V = \frac{V_{fwd} + V_{aft}}{2} \qquad \text{(B-7)}$$

$$= \frac{3.66 + 3.90}{2}$$

$$= 3.78 \text{ ft/sec}$$

$$S = \frac{S_{fwd} + S_{aft}}{2} \qquad \text{(B-8)}$$

$$= \frac{2.72 + 2.78}{2}$$

$$= 2.75 \text{ ft}$$

$$N = \frac{N_{fwd} + N_{aft}}{2} \qquad \text{(B-9)}$$

$$= \frac{80.6 + 84.2}{2}$$

$$= 82.4 \text{ steps/min}$$

The phase times and ratios are obtained from the stride relationships. The strides were selected from heel-strike to heel-strike. The subject completed two full strides by each foot in both the forward and aft directions.

For this run, the events of the first full stride of the right foot were recorded as

Phase	Frame
RHS	43
RTO	72
RHS	90

The total number of frames for this stride is

$$\begin{aligned} Fr_{st} &= Fr_{RHS_2} - Fr_{RHS_1} \\ &= 90 - 43 \\ &= 47 \end{aligned} \tag{B-10}$$

The number of frames during the swing is

$$\begin{aligned} Fr_{sw} &= Fr_{RHS_2} - Fr_{RTO} \\ &= 90 - 72 \\ &= 18 \end{aligned} \tag{B-11}$$

The number of frames during the support phase is

$$\begin{aligned} Fr_{sup} &= Fr_{st} - Fr_{sw} \\ &= 47 - 18 \\ &= 29 \end{aligned} \tag{B-12}$$

The same calculations were performed for the other seven strides. The results are compiled in tabular form:

	Right Leg				Left Leg				
	Forward		Aft		Forward		Aft		Sum
Stride Number	1	2	3	4	5	6	7	8	8
Stride Frames	47	47	47	44	48	46	44	45	368
Swing Frames	18	17	20	16	20	19	17	18	145
Support Frames	29	30	27	28	28	27	27	27	223

The average swing time is

$$t_{sw} = \frac{(0.03125)\ \Sigma \text{ swing frames}}{\text{number of strides}} \tag{B-13}$$

$$= \frac{(0.03125)\ 145}{8}$$

$$= 0.566 \text{ sec}$$

The average stride time is

$$t_{st} = \frac{(0.03125)\ \Sigma \text{ stride frames}}{\text{number of strides}} \tag{B-14}$$

$$= \frac{(0.03125)\ 368}{8}$$

$$= 1.437 \text{ sec}$$

The average support time is

$$t_{sup} = \frac{(0.03125)\ \Sigma \text{ support frames}}{\text{number of strides}} \tag{B-15}$$

$$= \frac{(0.03125)\ 223}{8}$$

$$= 0.871 \text{ sec}$$

The swing-to-support ratio is

$$\text{swing-to-support ratio} = \frac{t_{sw}}{t_{sup}} \tag{B-16}$$

$$= \frac{0.566}{0.871}$$

$$= 0.650$$

Table B-II

Summary of Reduced Data

Gravity Level	Vel.	Step Length	Step Freq.	Swing Time	Support Time	Stride Time	Swing to Stride Ratio	Swing to Support Ratio
	V	S	N	t_{sw}	t_{sup}	t_{st}	$\frac{t_{sw}}{t_{st}}$	$\frac{t_{sw}}{t_{sup}}$
(g)	(ft/sec)	(ft)	(steps/min)	(sec)	(sec)	(sec)	(%)	(%)
Subject A								
1.0	3.78	2.75	82.4	0.566	0.871	1.437	39.4	65.0
0.7	4.14	2.83	87.7	0.608	0.767	1.376	44.2	79.1
0.4	3.64	2.63	83.7	0.642	0.775	1.417	45.3	83.0
0.25	3.52	2.70	79.7	0.720	0.795	1.515	47.4	90.4
0.17	3.17	2.82	67.4	0.826	0.924	1.750	47.2	89.5
0.10	2.69	2.55	63.2	0.813	0.956	1.769	45.9	84.0
Subject B								
1.0	3.83	2.68	85.7	0.551	0.853	1.404	39.3	65.6
0.7	3.89	2.83	81.4	0.601	0.886	1.487	40.4	67.8
0.4	4.05	3.19	76.2	0.691	0.875	1.568	44.1	79.0
0.25	3.88	3.14	74.2	0.695	0.895	1.590	43.7	77.8
0.17	3.57	3.31	64.2	0.880	0.978	1.859	47.4	90.0
0.10	2.69	2.67	60.0	0.906	1.016	1.922	47.1	89.1

Appendix C

Sample Calculations in the Design of the Force-Measuring Walkway

Structural Design

Walkway Platform. The design condition for each walkway platform was that of a 200-lb man standing at mid-span of only one of the aluminum channels. In addition to the man's weight, one-half the weight of the platform was assumed to be uniformly distributed along its length.

The deflection of a simply supported beam with a concentrated load applied at the center, as determined by the Engineering Theory of Bending, is given by the expression

$$\delta_p = \frac{PL^3}{48\,EI} \qquad (C\text{-}1)$$

where

δ_p = maximum deflection, in.

P = concentrated load, lb

L = length of the beam, in.

E = modulus of elasticity, lb/in^2

I = moment of inertia of the cross-sectional area, in^4

The moment of inertia for a standard 4 x 1.580 x 0.180 in. channel is 3.83 in^4 (Ref 4:1-196). The modulus of elasticity for aluminum is 10×10^6 lb/in^2. For the given conditions, the deflection of the 12-ft beam due to the concentrated load at mid-span is determined to be

$$\delta_p = \frac{200\,(144)^3}{48\,(10 \times 10^6)(3.83)}$$

$$= 0.324 \text{ in.}$$

The mid-span deflection of a uniformly loaded, simply supported beam is given by the expression:

$$\delta_w = \frac{5}{384}\frac{WL^3}{EI} \qquad \text{(C-2)}$$

where

$W = wL$, lb

in which

w = uniform load per unit of length, lb/in.

The mid-span deflection due to a uniformly distributed load of 50 lb (one-half the assumed platform weight) is

$$\delta_w = \frac{5\,(50)(144)^3}{384\,(10 \times 10^6)\,3.83}$$

$$= 0.051 \text{ in.}$$

By superposition, the total mid-span deflection of the beam is

$$\delta = \delta_p + \delta_w \qquad \text{(C-3)}$$

$$= 0.324 + 0.051$$

$$= 0.375 \text{ in.}$$

Vertical Force Beam. The maximum allowable strain at the strain gage location is 0.002 in/in. for the maximum load condition. The end load is 225 lb. The distance between the point of application of the load and the strain gage location is 6 in.

The strain at the gage location may be computed by use of the formula

$$\epsilon = \frac{My}{EI} \qquad \text{(C-4)}$$

where

ϵ = strain, in/in.

M = bending moment, lb-in.

E = modulus of elasticity, lb/in^2

I = moment of inertia, in^4

y = distance from the neutral axis to the face of the beam, in.

For a beam of rectangular cross section, the moment of inertia is

$$I = \frac{bh^2}{12} \tag{C-5}$$

where

b = beam width, in.

h = beam height, in.

and

$$y = \frac{h}{2} \tag{C-6}$$

The moment is

$$M = PL \tag{C-7}$$

The bending strain at the extreme fibers of a cantilever beam of rectangular cross section may thus be expressed as

$$\epsilon = \frac{6PL}{Ebh^2} \tag{C-8}$$

The strain at the strain gage location for the selected 0.75 x 0.75 in. beam is therefore

$$\epsilon = \frac{6(225)(6)}{(10 \times 10^6)(0.75)(0.75)^2}$$

$$= 0.00192 \text{ in./in.}$$

The unit strain caused by a one-pound load may be computed by use of Eq C-8

$$\epsilon = \frac{6PL}{Ebh^2} \qquad (C\text{-}8)$$

However, it is more simply determined by dividing the total strain by the maximum load, that is

$$\epsilon_o = \frac{\epsilon}{P} \qquad (C\text{-}9)$$

where ϵ_o is the unit strain, inches per inch per pound.

The unit strain for the vertical force beam is therefore

$$\epsilon_o = \frac{0.00192}{225}$$

$$= 8.5 \times 10^{-6}$$

$$= 8.5 \text{ microinches per inch per pound}$$

The deflection of an end-loaded cantilever beam is given by the expression

$$\delta = \frac{PL^3}{3EI} \qquad (C\text{-}10)$$

which, for a beam of rectangular cross section, becomes

$$\delta = \frac{4PL^3}{Ebh^2} \qquad (C\text{-}11)$$

One inch, the distance between the strain gage and the change in cross-sectional area of the beam, is added to the beam length for the purpose of computing deflections; that is, L = 7 in. The deflection of the vertical force beam is therefore

$$\delta = \frac{4(225)(7)^3}{(10\times 10^6)(0.75)(0.75)^2}$$

$$= 0.073 \text{ in.}$$

<u>Horizontal Force Beam</u>. The design condition for the horizontal force beam is an end load of 50 lb. The distance between the point of application of the load and the location of the strain gage is 5 in. The dimensions of the selected beam are b = 0.50 in., and h = 0.40 in. The strain at the location of the strain gage is computed by the use of Eq C-8

$$\epsilon = \frac{6PL}{Ebh^2} \tag{C-8}$$

$$= \frac{6(50)(5)}{(10\times10^6)(0.50)(0.40)^2}$$

$$= 0.00188 \text{ in./in.}$$

By use of Eq C-9 the unit strain in the horizontal force beam is

$$\epsilon_o = \frac{\epsilon}{P} = \frac{0.00188}{50} = 0.0000376$$

or

$$\epsilon_o = 37.6 \text{ microinches per inch per pound}$$

The design length used to compute deflections is 5.625 in. The deflection is computed by use of Eq C-11

$$\delta = \frac{4PL^3}{Ebh^2} \tag{C-11}$$

The deflection of the horizontal force beam is therefore

$$\delta = \frac{4(50)(5.625)^3}{(10\times10^6)(0.50)(0.40)^2}$$

$$= 0.045 \text{ in.}$$

Strain Gage Instrumentation

The gage factor of a strain gage is the dimensionless ratio of the change in gage resistance to the change in strain, and may be expressed mathematically as (Ref 12:18)

$$F = \frac{\Delta R/R}{\epsilon} \qquad \text{(C-12)}$$

where

F = gage factor

R = nominal gage resistance, ohm

$\frac{\Delta R}{R}$ = change in resistance, ohm

ϵ = strain, in./in.

The voltage output of an unbalanced bridge circuit is given by the expression (Ref 12:61)

$$E_o = \frac{E R_G}{4(R+R_G)}\left(\frac{\Delta R_1}{R_1} - \frac{\Delta R_2}{R_2} + \frac{\Delta R_3}{R_3} - \frac{\Delta R_4}{R_4}\right) \qquad \text{(C-13)}$$

where

E_O = output voltage, v

E = applied voltage, v

R = nominal gage resistance, ohm

$\frac{\Delta R}{R}$ = change in resistance of any leg of the bridge circuit, ohm/ohm

R_G = galvanometer resistance, ohm

Vertical Force Circuit. The vertical force circuit consists of eight 350-ohm gages mounted two each on the four vertical force cantilever beams. The eight gages are connected to form a parallel bridge circuit with four active arms. Each arm has two gages connected in parallel.

Placing a unit strain on one beam, for example beam A_1, will cause a change in resistance of gages 1 and 2 of

$$\frac{\Delta R_1}{R_1} = -\frac{\Delta R_2}{R_2} = F\epsilon_0 \tag{C-14}$$

See Fig. 10 for gage numbers and locations.

The change in resistance of a leg containing n gages in parallel is

$$\frac{\Delta R_I}{R_I} = \frac{1}{n}\frac{\Delta R_i}{R_i} \tag{C-15}$$

where

$\frac{\Delta R_I}{R_I}$ = change in resistance of one leg

$\frac{\Delta R_i}{R_i}$ = change in resistance of any one gage

Therefore, the respective resistance changes in legs I and II are

$$\frac{\Delta R_I}{R_I} = -\frac{\Delta R_{II}}{R_{II}} = \frac{1}{2}\frac{\Delta R_1}{R_1} = \frac{F\epsilon_0}{2} \tag{C-16}$$

The output of the parallel bridge circuit for a unit strain in only one beam therefore becomes

$$E_0 = \frac{E R_G}{4(R+R_G)}\left(\frac{F\epsilon_0}{2} - \frac{(-)F\epsilon_0}{2} + 0 - 0\right)$$

$$= \frac{E\ R_G F\epsilon_0}{4\ (R+R_G)} \tag{C-17}$$

The input voltage is limited by the allowable current at any strain gage. To prevent overheating, the current at any gage is limited to 30 ma. The current at gage 1 of the parallel bridge circuit is equal to one-half the current flowing in leg I. Therefore, the allowable current in leg I is equal to twice the allowable current in gage 1, or 60 ma.

The allowable voltage across the parallel bridge circuit is therefore

$$E = I_I (R_I + R_{III}) \qquad (C\text{-}18)$$

$$= 0.60\ (175 + 175)$$

$$= 21 \text{ volts}$$

The galvanometer resistance is usually selected to be much greater than the nominal gage resistance, that is

$$R_G \gg R \qquad (C\text{-}19)$$

which reduces Eq C-17 to the approximate form

$$E_o \doteq \frac{E F \epsilon_o}{4} \qquad (C\text{-}20)$$

Assuming a gage factor of 2.0, and the unit strain of one vertical force beam equal to 8.6×10^{-6} in./in., the output of the vertical force circuit for a unit load applied to only one beam is computed by use of Eq C-20 to be

$$E_o \doteq \frac{E F \epsilon_o}{4} \qquad (C\text{-}20)$$

$$\doteq \frac{21\ (2.0)\ (8.6 \times 10^{-6})}{4}$$

$$\doteq 90 \times 10^{-6} \text{ v}$$

$$\doteq 90 \text{ microvolts per pound of load}$$

If the unit load is distributed equally or partially among the other beams and the respective values are substituted into Eq C-13, the same output will result. Thus, the output is independent of which beams in the circuit support the load. The equation

$$E_o \doteq \frac{E F \epsilon_o}{4} \qquad (C\text{-}20)$$

therefore applies, and ϵ_o can be considered to be the unit strain applied to the walkway platform, regardless of its distribution.

<u>Horizontal Force Circuit</u>. The horizontal force measuring circuit for each platform consists of four 350-ohm strain gages mounted on the horizontal force cantilever beam, as shown in Fig. 10. Any strain in the cantilever will activate all four bridge arms simultaneously. The output of the circuit is given by the equation

$$E_o = \frac{E F \epsilon R_G}{(R + R_G)} \tag{C-21}$$

which, by substitution of the basic assumptions stated for the vertical force circuit, becomes

$$E_o \doteq E F \epsilon_o \tag{C-22}$$

For equal gage resistances, the same voltage may be applied to a parallel bridge circuit regardless of the number of gages in parallel in each arm.

The unit strain of the horizontal force beams was determined to be 37.6×10^{-6} in./in. The corresponding output of the horizontal force measuring circuit for a unit load is computed by use of Eq C-22 to be

$$E_o \doteq E F \epsilon_o \tag{C-22}$$

$$= 21\ (2.0)\ (37.6 \times 10^{-6})$$

$$= 1580 \times 10^{-6}\ v$$

$$= 1.58 \text{ millivolts per pound of load}$$